QUANTIFYING MUSCULOSKELETAL RESERVE IN THE ELDERLY

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DECLARATION OF ORIGINALITY

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ABSTRACT

Developments in modern medicine, promotion of well-being in the elderly, and a greater understanding of healthy lifestyles have all culminated in our current 'ageing population'. It is predicted that by 2050 the population of over 65s will be 20% of the estimated general population. Most ageing research recognises that the ageing population has repercussions for both the economy and healthcare systems, however, this research fails to address the consequences of physiological changes in this ageing population.

Our bodies are not designed to perform at high levels of activity past certain ages. Muscle mass begins to decline around the 5th decade with diminishing internal systems, making activities which were easier at a younger age more strenuous on the body in later life. 'Musculoskeletal Reserve' (MSKR) is the concept of a reduced strength capacity in response to external stressor events, which is of greater relevance in the injured or severely frail. MSKR is quantifiable, and therefore can be used in conjunction with tools such as musculoskeletal modelling in approaching the problem of declining functional ability.

The movements of 95 eighteen to ninety-five year-olds were measured during a sit-to-stand task to better understand how the current elderly population is adapting to a lack of MSKR. This research was achieved through collaboration with the Royal Hospital Chelsea. This group had a significantly different muscle recruitment strategy when compared to the young group, which may have been protecting vulnerable joints. The younger group used contraction of the quadriceps to lift their body to stability from a quasi-static position after leaving the seat. The elderly subjects had much lower quadriceps forces, thus lowering the forces on the knee joint, and compensated by increasing the activity of the hamstring and calf muscles as an alternative in reaching the final standing position. The action of these compensatory muscle groups is less naturally efficient than those of the quadriceps, and despite significant decreases in joint force, places larger forces on muscles not usually recruited for such activities.

The changes in muscle recruitment were found to be subject to an alteration in movement strategy. The young made use of fast, tetanic contraction of the quadriceps to lift the body, whereas the elderly used prolonged activity of the alternative muscle groups listed above to stand from a seated position. This shift from acceleration to muscle effort makes simple, everyday activities more strenuous on the body. More importantly, it reduces the MSKR which constitutes the body's ability to overcome the external stressors of injury or MSK-related disease. Therefore, injury of any compensatory muscle will cause functional decline. This was most profound when rising without the assistance of arm-rest support, which when used would dramatically increase elderly joint forces in the shoulder when compared with the young; a potential intervention in shoulder postural position at the start of motion may reduce joint forces in this task. A premature reduction in lower limb strength with age places more reliance on the upper limb muscles to complete everyday activities, resulting in further lower limb atrophy through disuse. The eventual decline of upper limb strength leaves the elderly with a lack of MSKR to complete the activities, leading to increased vulnerability, dependence on others and risk of institutionalisation.

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GENERAL ABBREVIATIONS

Abbreviation	Full term
ADL	Activity of daily living
AP	Anterior-posterior
ASIS	Anterior superior iliac spine
BH	Body height
BMI	Body mass index
BSP	Body segment parameter
BW	Body weight
CGA	Comprehensive geriatric assessment
CoM	Centre of mass
CoP	Centre of pressure
EFS	Edmonton Frail Scale
EL	Elderly
EMG	Electromyography
FD	Forward dynamics
GH	Glenohumeral
GRF	Ground reaction force
ID	Inverse dynamics
JRF	Joint reaction force
MA	Middle-Aged
MFUB	Muscle force upper boundary
ML	Medio-lateral
MVC	Maximum voluntary contraction
NRMSE	Normalised root-mean-square-error
MSK	Musculoskeletal
MSKR	Musculoskeletal Reserve
OA	Osteoarthritis
PCSA	Physiological cross-sectional area
PF	Patellofemoral
SD	Standard deviation
STGP	Scapula-thoracic gliding plane
STS	Sit-to-stand
TF	Tibiofemoral
UKNSM	United Kingdom National Shoulder model
YO	Young

UPPER BODY MUSCLE ABBREVIATIONS

Abbreviation	Full name
SA	Serratus Anterior
Rmaj, Rmin	Rhomboids
LS	Levator Scapulae
Trap_scap, Trap_clav	Trapezius
SBS	Subscapularis
SS	Supraspinatus
IS	Infraspinatus
Tmin	Teres minor
Tmaj	Teres Major
Pmaj_clav, Pmaj_thor	Pectoralis Major
Pmin	Pectoralis Minor
DeltA, DeltM, DeltP	Deltoids (Anterior/Middle/Posterior)
BIC_short, BIC_long	Biceps brachii (Short/Long head)
CB	Coracobrachialis
Bra	Brachialis
TRI_long, TRI_med, TRI_lat	Triceps brachii (Long/Medial/Lateral)
LD	Latissimus Dorsi
Anc	Anconeous
BRD	Brachioradialis
SUP_hum	Supinator humerus
PT_hum	Pronator humerus

LOWER BODY MUSCLE ABBREVIATIONS

Abbreviation	Full name
Add. Brev	Adductor Brevis
Add. Long	Adductor Longus
Add. Magn	Adductor Magnus
Bic. fem. CL	Long head of Biceps Femoris
Bic. fem. CB	Short head of Biceps Femoris
Ext. dig. Long	Extensor Digitorum Longus
Ext. hal. Long	Extensor Hallucis Longus
Flex. dig. Long	Flexor Digitrum Longus
Flex. hal. Long	Flexor Hallucis Longus
Gastrocn	Gastrocnemius
Gemellus	Gemellus
Glut. Max	Gluteus Maximum
Glut. Med	Gluteus Medius
Glut. Min	Gluteus Minimus
Gracilis	Gracilis
Iliacus	Iliacus
Obt. Ext	External Obturator
Obturator int	Internal Obturator
Pectineus	Pectineus
Peroneus brev	Peroneus Brevis
Peroneus long	Peroneus Longus
Peroneus tert.	Peroneus Tertius
Piriformis	Piriformis
Plantaris	Plantaris
Popliteus	Popliteus
Psoas minor	Psoas Minor
Psoas major	Psoas Major
Quadratis fem	Quadratis Femoris
Rectus fem	Rectus Femoris
Sartorius	Sartorius
Semimembr	Semimembrinosis
Semitend	Semitendinosis
Soleus	Soleus
Tensor fasc. L	Tensor Fasciae Latae
Tibialis ant	Tibialis Anterior
Tibial post	Tibialis Posterior
Vastus interm	Vastus Intermedius
Vastus lat	Vastus Lateralis
Vastus med	Vastus Medialis

CHAPTER 1

INTRODUCTION

1.1 Scope of the thesis

The population has seen a dramatic increase in the number of people over the age of 65 years, leading to rising numbers of musculoskeletal conditions and subsequent dependence on others. This thesis will investigate musculoskeletal functional decline within a healthy ageing population for a key activity of daily living that is physically challenging and a suitable surrogate of many other musculoskeletal activities.

1.2 Motivation for thesis

Due to advancements in medicine, the prioritisation of improved global elderly well-being (United Nations. Department of Economic and Social Affairs. Population Division., 2013), and a much greater understanding of how to achieve a healthy lifestyle, the population of those over the age of 65 has dramatically increased over the last 100 years (Sander et al., 2014). It is predicted that by 2050 the elderly (individuals over the age of 65) population will reach 2 billion (WHO, 2011), 20% of the predicted numbers for the general population. The majority of research into ageing, in particular those with a focus on physiological changes in the elderly, emphasises the repercussions for both the economy and healthcare systems.

The past 40 years have shown a steadily increasing average life expectancy (Office of National Statistics, 2013). In response to this comes an increased demand to stay in employment longer. The number of people aged 65 or older still in employment has doubled in the last 20 years (Office of National Statistics, 2012). The retirement age is due to gradually rise to 68 by 2046 (Department of Work and Pensions, 2014), with recent recommendations to bring this forward to 2039 (Department of Work and Pensions, 2017). Therefore, larger numbers of the public will be expected to work for longer, and this is in addition to those who have a desire to remain employed and active beyond retirement. For those in the labour industry, or other jobs which require strenuous activity, there is an added risk of an increase in musculoskeletal (MSK) conditions which account for 41% of current work-related illnesses (Health and Safety executive, 2016). It is already the case that MSK-related conditions are one of the biggest surgical costs to the NHS (Department of Health, 2011) and affect around 25% of the elderly population (Wolff et al., 2002).

In a *Health Survey for England* report, it was found that upwards of 10% of over 65s struggled to complete the most basic of activities of daily living (ADLs) without assistance or incurring considerable discomfort (Whalley, 2012). Activities such as ascending/descending stairs, having a bath or shower, or simply dressing yourself proved too difficult to do comfortably or independently for 5 - 30 % of men and women, with approximately 1 - 5% unable to perform the activity at all.

To add further complication, the human body is not designed to perform at high levels of activity past certain age boundaries. Muscle volume and numbers of muscle fibres begin to decline as early as the 3rd decade, with the most noticeable changes occurring from the 5th decade onwards (Lexell et al., 1988), making activities which were easier at a younger age more strenuous on the body in later life. These changes, amongst others, have been well documented and the effects on strength and functional ability are known for the elderly population. However, little work has been done directly linking frailty, declines in strength and ADL performance at a physiological level. Advancements in computational modelling of muscles and joints give researchers a deeper understanding of how muscle activations drive movement and how contact forces in joints could be affecting performance. Furthermore, modelling different age groups has rarely been explored and so existing research looking to compare the young and elderly is limited in its understanding of biomechanical changes. It is therefore necessary to further explore the biomechanics of an ageing body, looking at changing MSK characteristics over time; doing so will generate a deeper understanding as to why the physical elements of frailty are affecting functional ability.

'Musculoskeletal Reserve' (MSKR, Chapter 3), is a concept based on the theoretical levels of physical strength a person has beyond that of normal function. Introduced by Bull et al. (2008), MSKR describes the excess capability required to perform an ADL in spite of injury or muscle impairment. By essentially 'switching off' chosen muscle groups through the use of computational MSK models, it was found that a number of tasks were theoretically unachievable, thus suggesting a reduced MSKR. In consideration of how MSKR relates to an ageing population, Samuel et al. (2013) found that functional demand increased dramatically for the elderly during tasks such as ascending stairs, leaving minimal extra strength to cope with unsuspected falls or injuries. Considering the changes to the MSK system (Chapter 2) as the body ages, this reserve of strength will diminish. Muscle weakness is an important pre-requisite and indicator of a lower MSKR, where preliminary modelling tests show both the local effects of individual muscle weakness and indicate muscles with vital roles during ADLs (Chapter 6).

It is important for the elderly to maintain mobility in order to stay independent but many can become isolated from society if they are unable to leave their homes

without assistance. Addressing the biomechanical compensatory effects of ageing would not only help the elderly perform important ADLs around the home, but could also, through improved mobility, increase social interaction.

1.3 Objectives

The main objective of the thesis is therefore to increase understanding of how changes in muscular strength, and movement strategy, can effect ADL performance in different age groups. This lack of strength, or reduced MSKR, will be quantified and used in understanding why physiological changes are detrimental to long-term functional ability, and how difficulties in performing everyday activities can be addressed in a clinical setting.

1.4 Aims

These objectives will be achieved by:

- reviewing existing literature on the changing musculoskeletal characteristics of an ageing population, and how degradation of the body can lead to a reduction in strength in the elderly;
- developing a tool for use in understanding how the concept of a reduced musculoskeletal reserve can alter the muscle recruitment strategy as part of the ageing process; and
- a combination of optical motion capture and musculoskeletal modelling, to be used in a study observing the movements of different age groups. The results will be used to better understand how the elderly have adapted to the previously investigated changes in the MSK system.

1.5 Thesis Structure

Chapter 2 lists the anatomical terminology used throughout the thesis and describes the local anatomy relevant to the work presented. This chapter also contains a review of the changes which occur to the musculoskeletal system with age, and the consequences on muscular strength.

Chapter 3 introduces frailty, and how it is currently assessed in a clinical setting. Having explored both muscular changes and frailty, the concept of musculoskeletal reserve is introduced which forms the main focus of the thesis. Chapter 3 finishes with a review of activities of daily living, focusing on the importance of the sit-to-stand motion.

Chapter 4 is a brief overview of the computational simulations used as data processing tools in this thesis. This includes models for both the upper and lower limbs, allowing full-body analysis of movement. The required inputs, processing steps and outputs are explained.

In addition to the research carried out in the thesis, Chapter 5 covers work looking at creating bespoke body segment parameters for use in the aforementioned musculoskeletal models. After a review of existing technology and use of regression tables, the proposed set up and data-processing protocols are covered. The results are compared with literature cadaveric work and geometric modelling. Group-specific regression tables are explored, and the improvements they could present for different populations are discussed.

Chapter 6 simulates muscle weakness, and explores how essentially 'turning off' certain muscles within the simulations described in Chapter 4, can have significant effects on joint contact force, neighbouring muscles' activation levels and the success of mathematical optimisation to solve the loadsharing redundancy during a sit-to-stand activity.

Chapter 7 is a comprehensive, biomechanical analysis of sit-to-stand for three different age groups of male participants. Differences in both upper and lower body

external forces, joint reaction forces and muscle activity are assessed for all ages and any significant differences are discussed.

Chapter 8 follows on directly from Chapter 7 with a further exploration of the biomechanical data. A discussion on the kinematics behind the motions of each age group is presented to explain the differences seen between age groups in terms of muscle activity and required joint torques. This chapter then moves onto the relevance of musculoskeletal reserve, and how reduced strength in the elderly forces changes in strategy that are not necessarily beneficial to an ageing body, despite successful completion of an ADL.

Chapter 9 is the concluding chapter that summarises the thesis findings and presents possibilities for future work. The original objectives and aims are addressed, and clinical applications for such work are discussed.

CHAPTER 2

THE ANATOMY OF AN AGEING MUSCULOSKELETAL SYSTEM

This chapter gives an overview of the musculoskeletal (MSK) system as relevant to the shoulder and lower limbs, in the context of an ageing MSK system.

2.1 The Musculoskeletal system

The MSK system comprises two main elements: the skeletal system and the muscular system. Both are pivotal in the safe and stable movement of the body.

2.1.1 The skeletal system

The skeletal system protects internal organs, provides anchorage for muscle attachment sites and facilitates rigid kinematics (Nordin & Frankel, 2001). Bones are connected at joints via ligaments, which act as passive stabilisers during movement. Bone structure changes with age, due to a reduction in bone mineral density (Nordin & Frankel, 2001) and higher prevalence of osteoporosis in the elderly (NIH, 2001). Despite these factors being present in all age groups, it can mean a higher susceptibility to injury in the elderly compared to younger groups (Greenspan et al., 1998), as bone tends to weaken with age (Ruff & Hayes, 1982) and is less able to withstand the high forces associated with, for example, the sudden impact of a fall.

This thesis focuses on the muscle activity at the shoulder and in the lower limb and therefore only requires knowledge of the relevant skeletal structures of these extremities. The key structures to be considered are: the Humerus, Scapula, Thorax, Clavicle, Ulna and Radius of the upper body, and Pelvis, Femur, Tibia, Fibula and Patella of the lower body.

2.1.2 The muscular system

The body's muscular system can be subdivided into three types of muscle: cardiac (composes the heart), smooth (lines the internal organs) and skeletal (which attaches to the skeleton via tendons) (Nordin & Frankel, 2001). Skeletal muscle is pertinent to this study as this type of muscle drives movement. Muscles are made up of a series of structures which vary in size. The largest structure is a muscle fibre, which consists of myofibrils containing contractile elements called Sarcomeres (Perry, 1992; Nordin & Frankel, 2001). It is the action of these elements that allow muscles to contract. Motor neurons (the nerves that transmit electrical signals from the brain), stimulate the muscle fibre forming a motor unit (Nordin & Frankel, 2001). This begins

the process of muscular contraction. Understanding this process and the factors which affect the strength of the contraction, help to understand why changes due to the ageing process can weaken the body.

2.1.2.1 Muscular contractions

A muscular contraction is the result of the two mutually repellent protein filaments which make up a sarcomere, Actin and Myosin (Figure 2.1), sliding over each other in opposing directions. When calcium (released into the sarcoplasmic reticulum surrounding the sarcomere) binds to a protein on the filament surfaces, cross-bridges are formed which facilitates the sliding action (Nordin & Frankel, 2001; Cram et al., 1998). The strength and change in length of the resulting muscular contraction depends on the number of motor units recruited, packing of muscle fibres both in parallel and series (Wisdom et al., 2014; Nordin & Frankel, 2001) and fibre types.

Muscle fibres are categorised into two forms: Type 1 ('slow-twitch') and Type 2 ('fast-twitch'), which can be further categorised into Type 2a and 2b (Wisdom et al., 2014; Ricoy et al., 1998). Each fibre type produces varying levels of maximal tension and fatigues at different rates. Type 1 is a low fatigue muscle, due to a high number of mitochondria and oxygen supply for subsequent efficient energy production (Nordin & Frankel, 2001). Type 1 is therefore more efficient for prolonged, low-intensity work such as maintaining posture. Type 2 produces high amounts of tension but is more easily fatigued (Nordin & Frankel, 2001). The additional categorisations 2a and 2b describe two variations on this structure. Type 2a has a good blood supply and can maintain contractile activity for longer periods of time, however, at high activity rates respiration exceeds energy production and the muscle will start to fatigue (Nordin & Frankel, 2001). Type 2b on the other hand has a low supply of neighbouring capillaries and so must respire anaerobically, but still produces energy rapidly and fatigues easily (Nordin & Frankel, 2001). These fibres can produce a relatively large amount of tension compared to Type 1 but only for short periods of time.

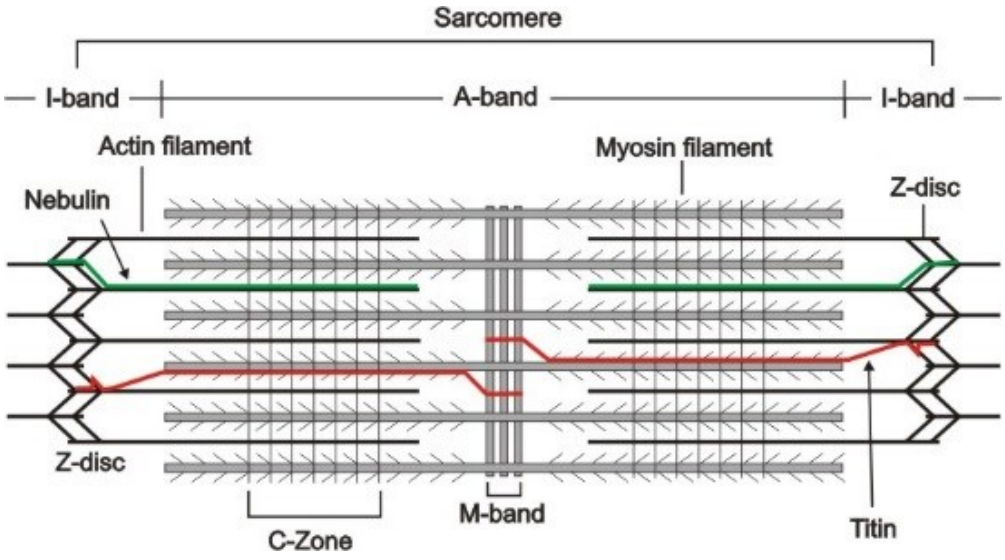


Figure 2.1 - A simplified image of the sarcomere structure. Reproduced with permission from Luther (2014)

Fibre type distributions will determine the contractile properties of individual muscles, and as a consequence, the body's functional abilities. Ricoy et al. (1998) found that within the Vastus Lateralis muscle, an important knee extensor, the largest percentages of Type 1 fibres were found in long distance runners and the smallest was found within those conducting high-intensity activities such as karate or triple jump. Type 2 fibres were more prevalent in swimmers, whereas footballers had the lowest percentage of these fibres. The effects of age on fibre type number and distributions are discussed in more detail in Section 2.4.3, but it is worth noting that research suggests fast-twitch muscles are more vulnerable to atrophy in later years (Lexell et al., 1983; 1988).

There are a number of relationships and properties which define the force production of any muscle during contractions. The length-tension relationship determines the amount of muscle force relative to the length of contraction, as shown in Figure 2.2. This additionally describes the relationship between length and a combination of both active and passive tension. Active tension describes the action of the contractile elements of a muscle, namely the previously described sarcomeres. Passive tension is developed when the contraction surpasses its resting length (point of maximum active tension) and the non-contractile elements (fatty or connective tissues)

are put under tension (Nordin & Frankel, 2001). Figure 2.2 shows that as the muscle length increases, the active tension also increases to a maximum point, but past this point it would decrease, if it were not for the passive elements which contribute to the overall muscle tension.

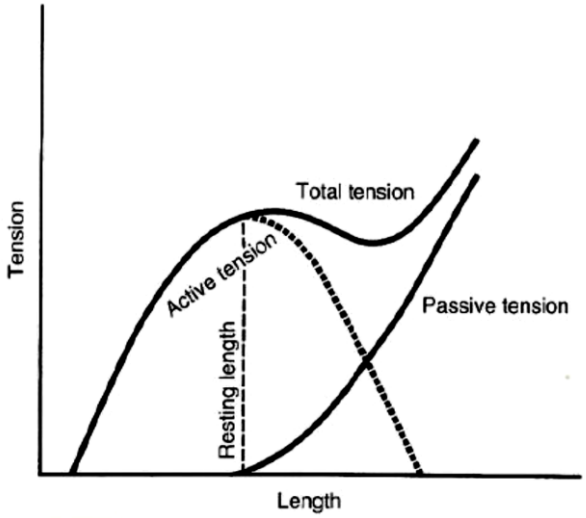


Figure 2.2 - The active/passive tension curve when a whole muscle contracts isometrically and tetanically, plotted against muscle length. Image reproduced, with permission, from Nordin & Frankel (2001)

The load-velocity relationship describes the speed of muscular contractions, which varies depending on the type of contraction: eccentric (lengthening) or concentric (shortening). Figure 2.3 shows how this relationship changes with both contraction types. In summary, the graph describes how zero-to-low loads produce concentric, fast contractions which slowly decrease in velocity as the load increases. Once at the point of maximum muscle force, the muscle contracts isometrically (no movement) but above maximum capacity, the muscle will elongate during contraction (Nordin & Frankel, 2001).

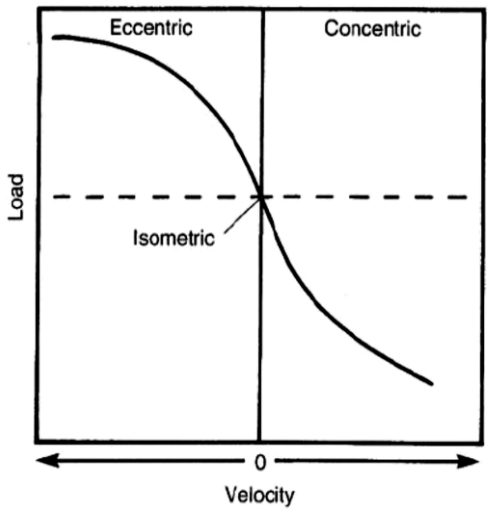


Figure 2.3 - Graph showing the load-velocity curve of muscular contractions. This was generated by plotting velocity of motion of the muscle level arm against external load. Image reproduced, with permission, from Nordin & Frankel (2001)

Each of these relationships is important in determining a muscle's ability to produce tension. Muscle architecture plays a pivotal role in these relationships and the changing aspects with age will be explored in later sections. The configuration of sarcomeres has a profound effect on the contractile properties, with both series and parallel layouts effecting different relationships. The more sarcomeres packed in series, the greater the myofibril length and ability to contract concentrically at higher velocities (Wisdom et al., 2014; Nordin & Frankel, 2001). Alternatively, a parallel-heavy configuration (or increased cross-sectional area) will increase the amount of active tension the muscle can produce, and therefore increases the peak of the length-tension curve (Wisdom et al., 2014; Nordin & Frankel, 2001). A decrease of muscle size, as found in numerous ageing studies (discussed in Section 2.4.2), will therefore undoubtedly have a negative effect on the ability of muscle to produce tension.

2.2 Anatomical Terminology

Motion of the human body is described using agreed upon clinical terminology, which is used consistently throughout this thesis. There are three main planes of movement: Sagittal, Transverse and Coronal (also described as Frontal), which can be

used to describe the whole body or segment orientation (Figure 2.4). Motions of the upper and lower limbs will also be described using clinical terminology (Figure 2.5).

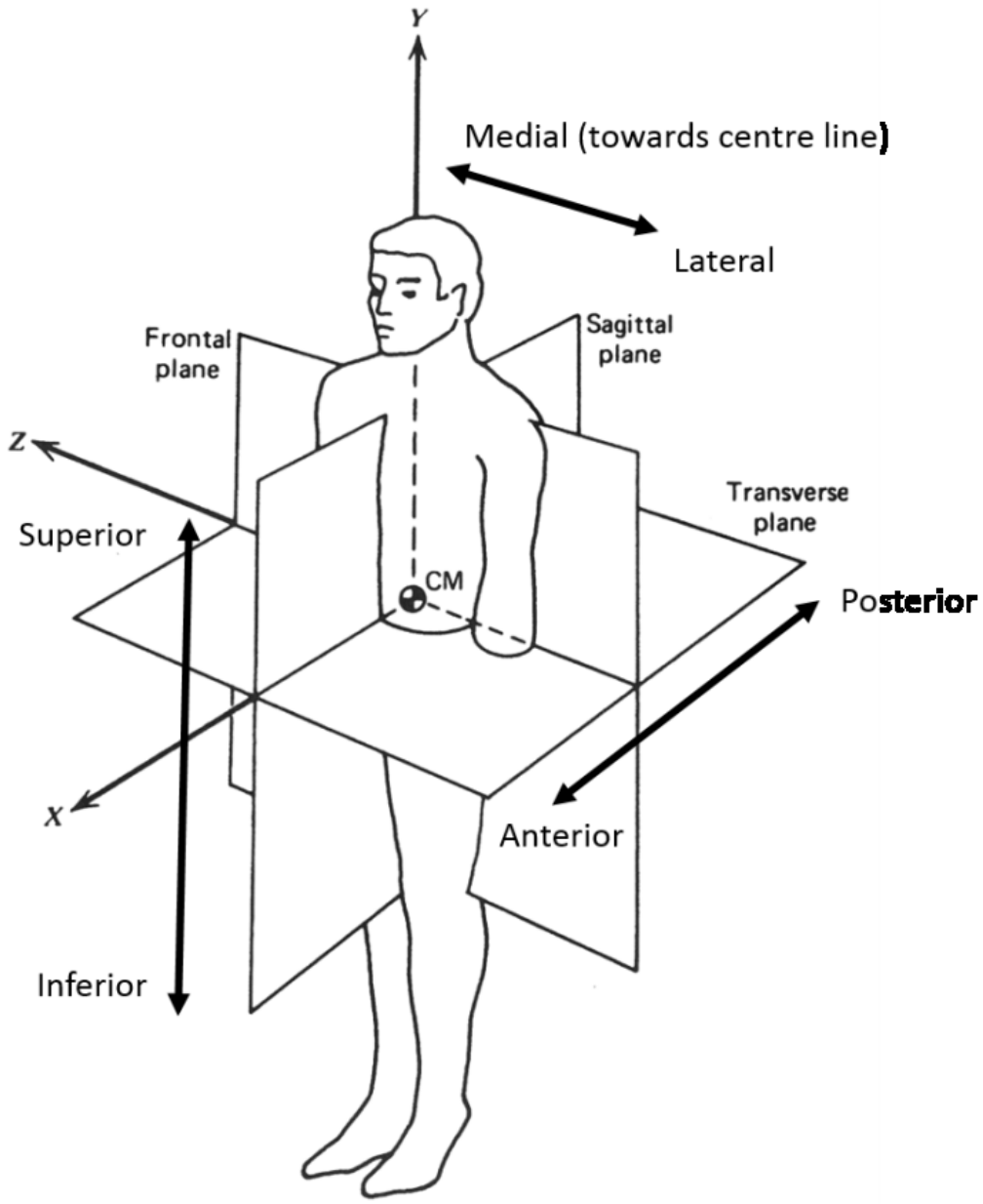


Figure 2.4 – The three main planes used to describe body motion. Also shown are the main anatomical directions used to describe parts of the body in relation to landmarks. 'Frontal' is the term used here to describe the Coronal plane. CM denotes the body's centre of mass. Image reproduced, and modified, with permission from Winter (2009)

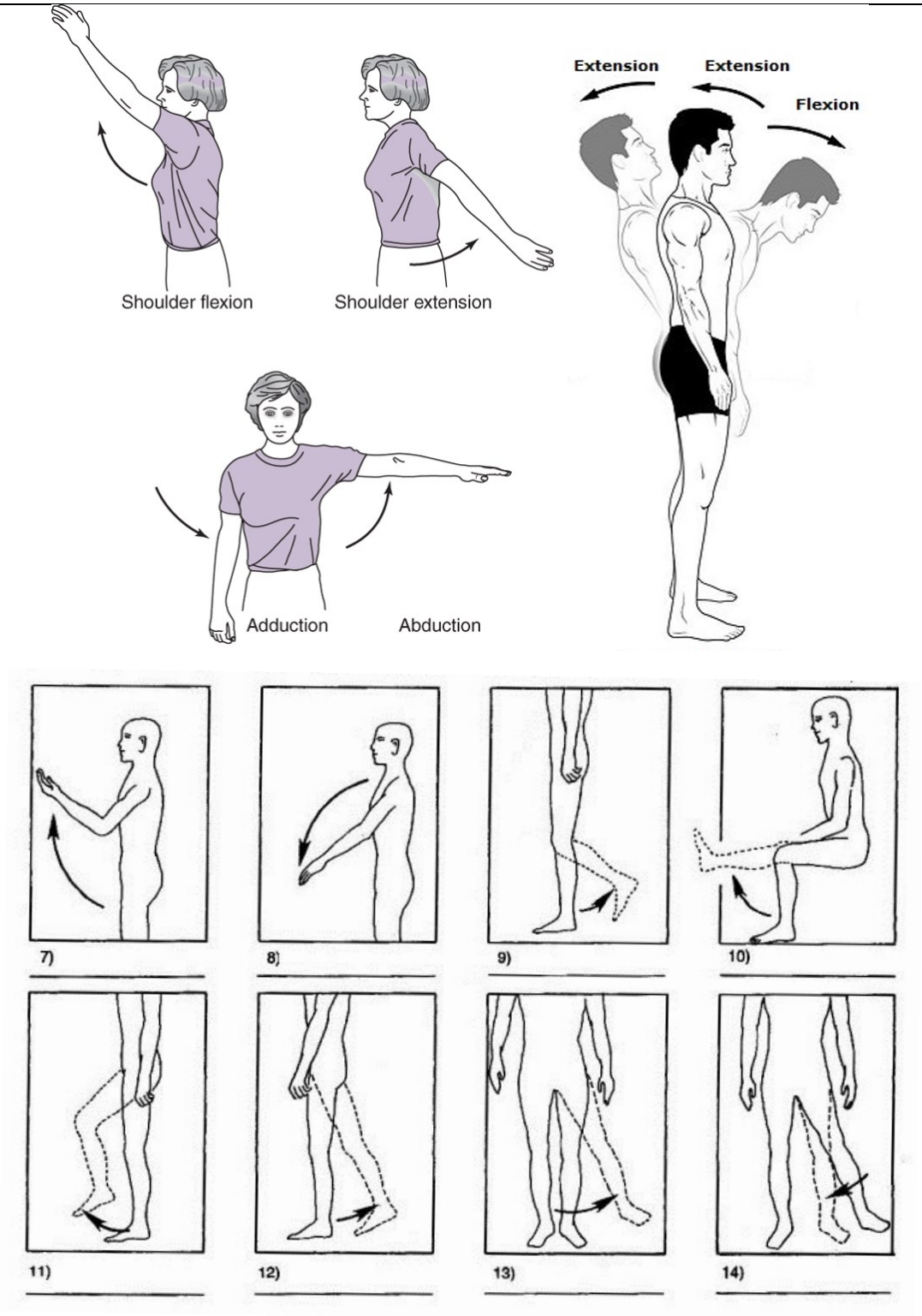


Figure 2.5 – Clinical terminology for upper and lower body motions. Shoulder movement images taken, and modified, from Skinner & McMahon (2014) with permission. Trunk movement taken from GetMedEdu (2017). Unlabelled images: 7 – elbow flexion, 8 – elbow extension, 9 – knee flexion, 10 – knee extension, 11 – hip flexion, 12 – hip extension, 13 – hip abduction and 14 – hip adduction. Lower image taken and modified from Maurer (2017)

It is important to define the motions of the scapula and clavicle, as part of this thesis will focus on the movements of the shoulder. The shoulder has a large range of motion in multiple directions (Figure 2.6 and Figure 2.7) due to a lack of bony articulation between the thorax and scapula (Paine & Voight, 2013), making it an important structure of the body when considering complex movements of the upper limbs. For instance, the large angle of abduction available to a human shoulder is only made possible by the upward rotation of the scapula. It is due to this rotational assistance that is it useful to describe humeral joint angles relative to the scapula anatomical frame.

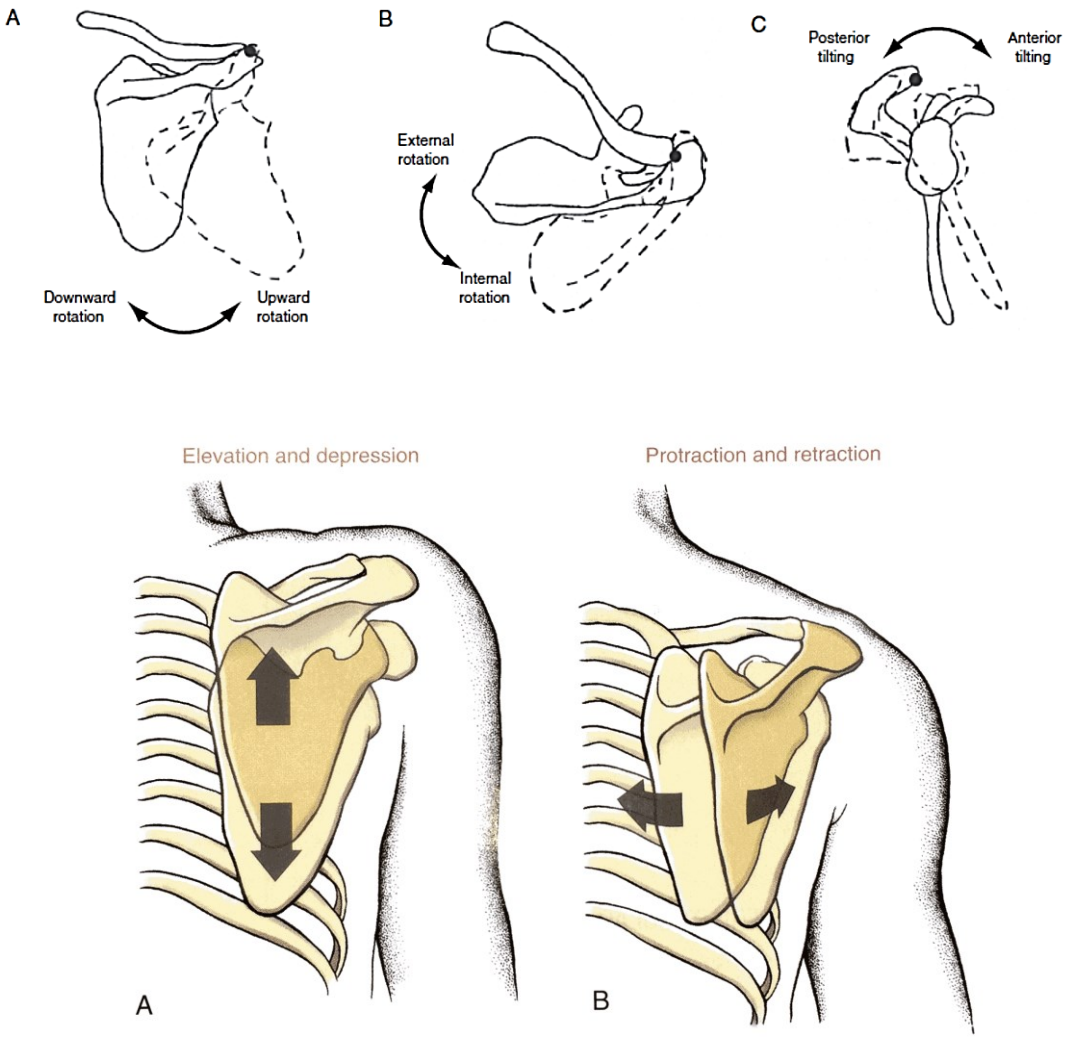


Figure 2.6 – Scapula rotations and translations. Top row: A - Posterior view, B - Superior view and C - Lateral view. Rotational axes indicated by a black dot. Image reproduced from Borich et al. (2006) with permission. Bottom row: A & B – Posterior views where elevation is superior, depression is inferior, retraction is medial, and protraction is lateral. Image reproduced from Neumann (2010) with permission

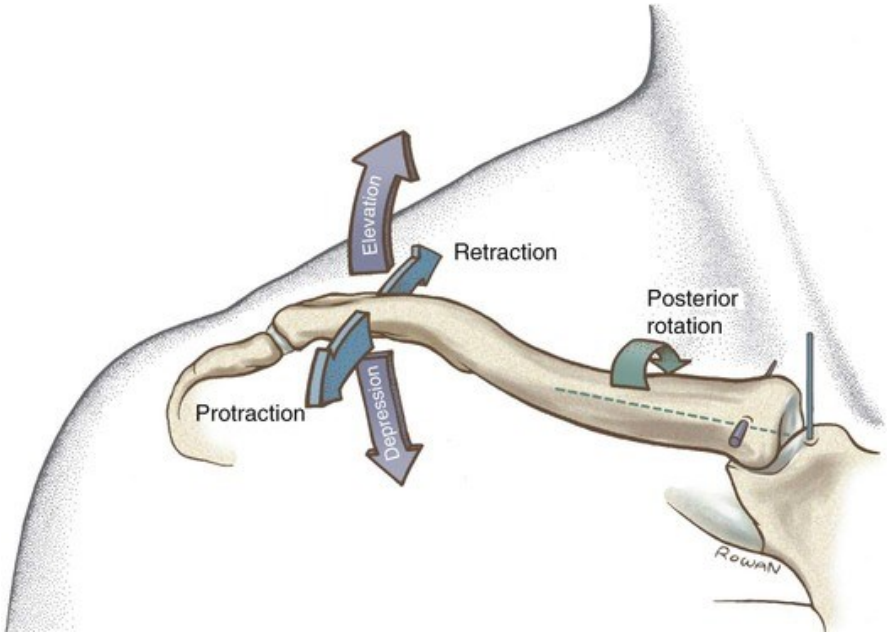


Figure 2.7 – Clavicle rotations. Image reproduced, with permission, from Neumann (2010)

2.3 Anatomy of the body

This thesis will look at the movements of the entire body, as the ageing process has a global effect on motion as opposed to injury or joint pathology which may only affect localised areas. It is therefore necessary to briefly introduce some specific regional anatomy.

2.3.1 Upper limb anatomy

The shoulder complex consists of the clavicle, scapula and proximal end of the humerus, as shown in Figure 2.8. The arm also consists of the ulna (medial side of the forearm), radius (lateral side of the forearm) and the hand bones, which are a large number of small bones termed carpals, metacarpals and phalanges. However, for the purposes of this thesis, the hand is considered as a single segment.

The clavicle is the only bony attachment between the trunk and the upper limb and serves as the origin for multiple muscles which connect to the thorax, neck and

upper limb (Drake et al., 2010). The clavicle also serves as an attachment point for three ligaments: the conoid, trapezoid and costoclavicular, all of which play important roles in constraining its movement (Pronk, 1991; Cave, 1961). The scapula plays a vital role in both shoulder movement and stability, details of subject-specific tracking of which can be found in Chapter 7. The scapula is a large flat bone, as shown in Figures 2.8 and 2.9, which sits on the superior, posterior region of the thorax, on one side of the spine. There are no bony attachments between the scapula and thorax (Paine & Voight, 2013), allowing the shoulder to have multiple degrees-of-freedom. The scapula moves relative to the thorax along the scapula-thoracic-gliding-plane (STGP), which holds the scapula close to the thorax during movement (Culham & Peat, 1993). Articulation of the humerus about the scapula is made possible by the glenoid cavity, which forms the glenohumeral joint with the proximal end of the humerus bone.

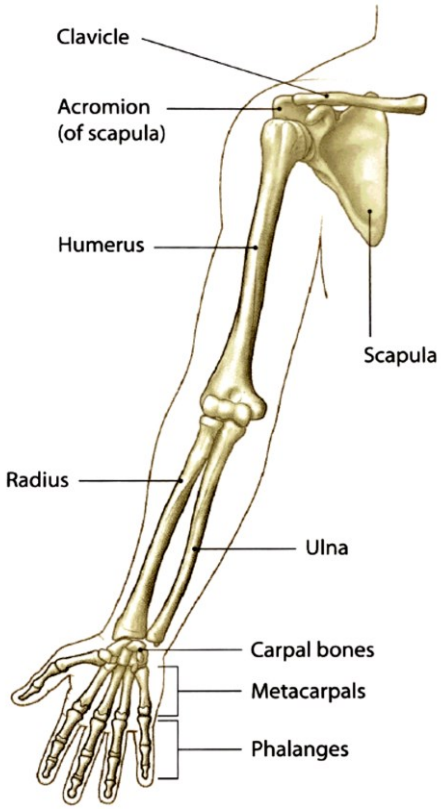


Figure 2.8 - Skeletal anatomy of the upper limb. Image reproduced with permission from Moses et al. (2013)

The upper limb musculature is listed in Table 2.1.The main stabilisers of the scapula include the serratus anterior, rhomboids, levator scapulae and trapezius, whereas

the glenohumeral joint is stabilised by the rotator cuff muscles: subscapularis, teres minor, supraspinatus and infraspinatus.

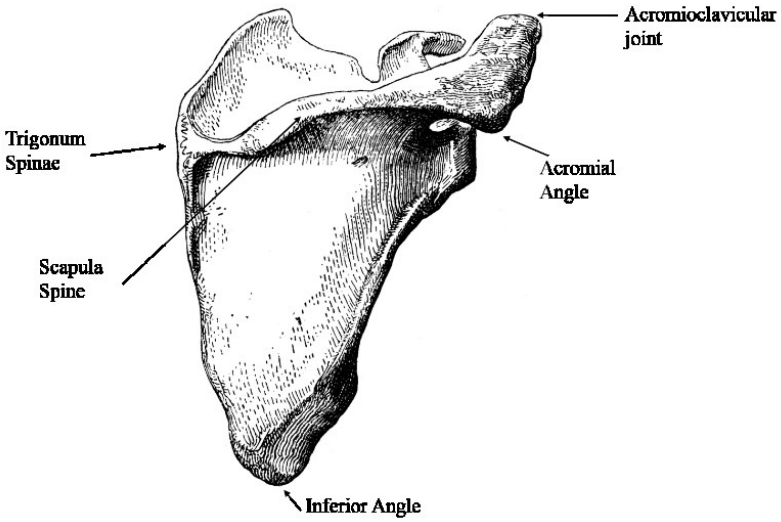


Figure 2.9 - Posterior surface of the scapula (Blaisdell, 1904)

2.3.1.1 Upper limb joints

The upper limb consists of three main joints: the shoulder complex (Figure 2.10; comprising the glenohumeral (GH), acromioclavicular, sterno-clavicular and scapulo-thoracic joints), the elbow and the wrist.

2.3.2 Lower limb anatomy

The lower body consists of four main articulating segments: the pelvis, thigh, shank and foot. The lower limb musculature and bones are shown in Table 2.2 (see Klein Horsman et al., 2007, for origins and insertions) and Figure 2.11 respectively. One of the important functions of the lower limb is to support body weight with minimal energy expenditure (Drake et al., 2010; Masouros et al., 2010). The secondary function is movement, with the appropriate terminology used in this thesis set out in Figure 2.5.

Table 2.1 - Musculature of the upper limb (Paine & Voight, 2013; Drake et al., 2010). Ant = Anterior, Med = Medial, Post = Posterior, Sup = Superior, Lat = Lateral and Inf = Inferior

Muscle	Origin	Insertion	Function
Serratus Anterior	First 8 ribs	Ant, Med scapula border	Stabilises scapula during elevation, pulls scapula forward and around thorax.
Rhomboids	Minor –C7 and T1, Major – T2 to T5	Med scapula border	Stabilises the Med scapula border.
Levator Scapulae	C1 to C4	Med scapula at scapular spine	Elevation and retraction
Trapezius	Sup nuchal line, external occipital protuberance, nuchal ligament, C7 to T12	Lat clavicle, acromion and scapula spine.	Scapula elevator, rotates humerus in abduction above horizontal. Middle fibres retract the scapula, and lower fibres depress.
Subscapularis	Med two-thirds of subscapular fossa	Lesser tubercle of the humerus	Rotator cuff. Med rotation of arm
Supraspinatus	Med two-thirds of scapula supraspinous fossa	Humerus greater tubercle Sup facet	Rotator cuff. Initiation of abduction to 15°
Infraspinatus	Med two-thirds of scapula infraspinous fossa	Middle facet on Post surface of humerus greater tubercle	Rotator cuff. Lateral rotation of arm.
Teres minor	Post scapula adjacent to Lat border	Inf facet on the Post surface of humerus greater tubercle	Rotator cuff. Lateral rotation of arm.
Teres Major	Post surface of the Inf scapula angle	Med lip of humerus Ant surface	Med rotation and extension of arm at shoulder
Pectoralis Major	Ant surfaces of the Med half of clavicle and Sternum	Lat lip of intertubercular sulcus of humerus	Arm adductor, flexor and medial rotator
Pectoralis Minor	Ant surfaces of 3rd, 4th and 5th ribs, related intercostal spaces	Scapula coracoid process	Depresses tip of the shoulder, scapula protractor
Deltoids (Ant, middle, Post)	Inf edge of scapula spine, Lat margin of acromion, Ant border of the Lat clavicle	Deltoid tuberosity of humerus	Arm abductor (past 15° facilitated by Supraspinatus), clavicle/Post fibres assist arm flexion and extension respectively
Biceps brachii (long/short head)	Supraglenoid scapula tubercle (L), coracoid process (S)	Radial tuberosity	Forearm flexor at elbow, also supinator for forearm
Coracobrachialis	Apex of coracoid process	Humerus mid-shaft, Med side	Arm flexor at shoulder
Brachialis	Ant humerus (Med and Lat), intermuscular septae	Tuberosity of ulna	Forearm flexor at elbow
Triceps brachii (Long/Med/ Lat)	Infraglenoid scapula tubercle (L); Post humerus (M/La)	Olecranon	Forearm extensor elbow, extend/ adduct shoulder (L only)
Latissimus Dorsi	T7 to L5 and sacrum, iliac crest and ribs 10 to 12	Intertubercular humerus sulcus	Humerus extensor, adductor and medial rotator
Anconeous	Lat epicondyle of humerus	Olecranon, Post ulna	Abduction of ulna in pronation, accessory extensor of elbow joint
Brachioradialis	Proximal Lat supraepicondylar humerus ridge, intermuscular septum	Lat surface of distal radius	Accessory flexor of elbow joint when forearm is mid-pronated
Supinator humerus	Supinator crest of ulna and Lat humerus epicondyle	Radius superior to the oblique line	Hand/Forearm supination
Pronator humerus	Med humerus epicondyle	Lat radius, midway along shaft	Hand/Forearm pronation

Table 2.2 - Musculature of the lower limb (Platzer, 2009; Drake et al., 2010)

Muscle	Function	Muscle	Function
Adductor Brevis	Adducts thigh at hip	Peroneus Longus	Foot plantarflexion
Adductor Longus	Adducts and medially rotates thigh at hip	Peroneus Tertius	Foot eversion
Adductor Magnus	Adducts and medially rotates thigh at hip	Piriformis	Laterally rotates femur at hip, abducts flexed femur at hip
Biceps Femoris (Long/Short)	Hamstring. Flexes leg at knee (L/S), extends and laterally rotates thigh at hip (L), laterally rotates leg at knee when flexed (L/S)	Plantaris	Foot plantarflexion and knee flexion
Extensor Digitorum Longus	Extension of lateral 4 toes and foot dorsiflexion	Popliteus	Knee stabiliser (resists lateral rotation of tibia on femur)
Extensor Hallucis Longus	Great toe extensor and foot dorsiflexion	Psoas Minor	Weak flexion of lumbar vertebral column
Flexor Digitrum Longus	Flexion of lateral 4 toes	Psoas Major	Flexes thigh at hip
Flexor Hallucis Longus	Great toe flexor	Quadratis Femoris	Laterally rotates femur at hip
Gastrocnemius	Foot plantarflexion and knee flexion	Rectus Femoris	Flexes thigh at hip, extends leg at knee
Gemellus	Laterally rotates extended femur at hip, abducts flexed femur at hip	Sartorius	Flexes thigh at hip, flexes leg at knee
Gluteus Maximum	Extends flexed femur at hip, lateral stabiliser of hip and knee, laterally rotates and abducts thigh	Semimembranosus	Hamstring. Flexes leg at knee, extends thigh at hip, medially rotates thigh at hip and leg at knee
Gluteus Medius	Abducts femur at hip, medially rotates thigh	Semitendinosis	As above
Gluteus Minimus	Abducts femur at hip, medially rotates thigh	Soleus	Foot plantarflexion
Gracilis	Adducts thigh at hip and flexes leg at knee	Tensor Fasciae Latae	Stabilises knee in extension
Iliacus	Flexes thigh at hip	Tibialis Anterior	Foot dorsiflexor, foot inversion
External Obturator	Laterally rotates thigh at hip	Tibialis Posterior	Foot plantarflexion and inversion
Internal Obturator	Laterally rotates extended femur at hip, abducts flexed femur at hip	Vastus Intermedius	Extends leg at knee
Pectineus	Adducts and flexes thigh at hip	Vastus Lateralis	Extends leg at knee
Peroneus Brevis	Foot plantarflexion	Vastus Medialis	Extends leg at knee

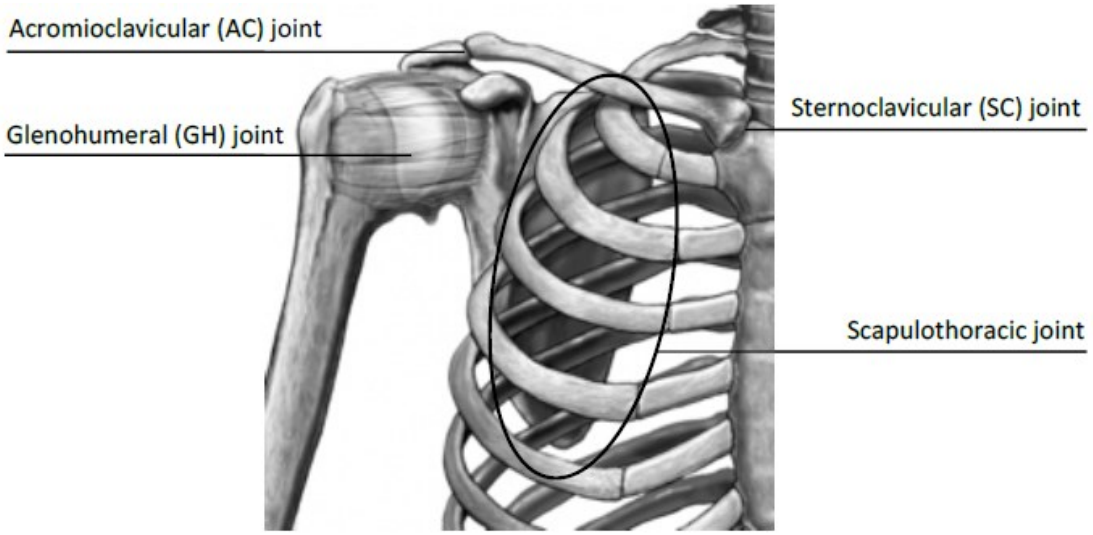


Figure 2.10 - Shoulder joint complex (Nucleus Medical Media, 2015)

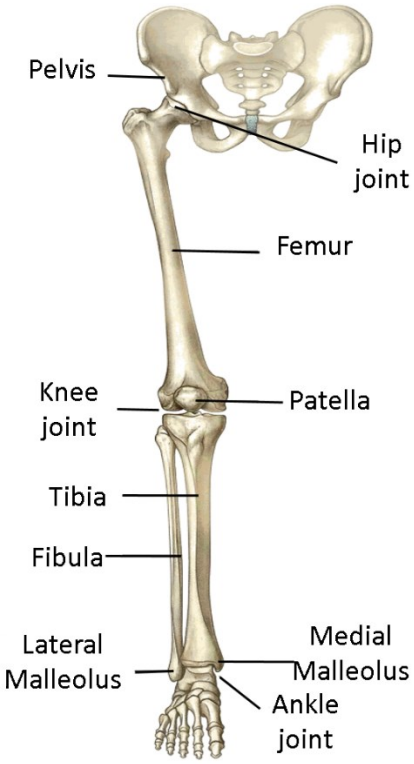


Figure 2.11 - Skeletal anatomy of the lower limb. Image reproduced and modified, with permission, from Drake et al. (2010)

2.3.2.1 Hip

The pelvis and femur make up the gluteal region of the lower limb, with a ball and socket joint (the hip joint) making up the junction between the pelvis and leg (Drake et al., 2010). This socket provides intrinsic strength to the joint, making it the most stable in the body (Nordin & Frankel, 2001). The main muscles facilitating movement about the hip include the gluteal muscles, rectus femoris, and hamstrings.

2.3.2.2 Knee

The knee joint consists of the link between the femur and tibial plateau, and plays a vital role in most daily activities which involve whole body movements. The role played by the knee in transmission of high loads and moments is important for safe and effective movement, but also makes it highly susceptible to injury (Masouros et al., 2010; Nordin & Frankel, 2001). There is a very high incidence of osteoarthritis in the knee joint, especially in those over the age of 80 (Dulay et al., 2015), which severely limits the range of motion and reduces activity levels in old age. The knee joint can be subdivided into two distinct joints: the tibiofemoral (TF) and patellofemoral (PF) joints, which describe the movement of the femur over the tibia, and patella over the femur, respectively. Both joints are of interest when considering high loads through the knee joint, but it is worth remembering that the muscular attachments of the lower limb will have a significant effect on shared-joint loading. For example, knee extension is facilitated by the quadriceps which insert onto the patella (Drake et al., 2010).

2.3.2.3 Ankle

The ankle joint is highly complex, and consists of three joints: the tibiotalar, fibulotalar and tibiofibular (Nordin & Frankel, 2001). Movement of the ankle is described as mostly hinge-like, allowing dorsi- and plantarflexion (Figure 2.12, Drake et al., 2010), however the joint can also produce abduction/adduction and eversion/inversion movements (Nordin & Frankel, 2001). The predominant extrinsic muscles which facilitate movement of the feet include the tibialis anterior, gastrocnemius, soleus and plantaris. For the sake of this thesis, the foot is considered as a single segment.

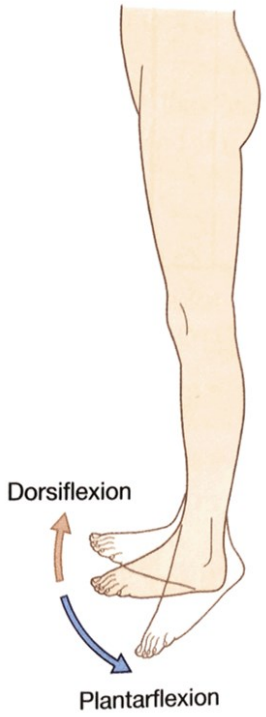


Figure 2.12 – Clinical terminology describing rotations of the ankle joint. Image reproduced with permission from Drake et al. (2010)

2.4 The ageing musculoskeletal system

2.4.1 Introduction

Throughout human life, the body adapts and remodels elements of the MSK system in order to provide sufficient strength and stability. Bone growth and development is fairly linear up until the age of 20 to 25 (a few years earlier in women - Hägg & Taranger, 1980; Mendes et al., 2010) after which bones are generally considered at full maturity (Drake et al., 2010). Puberty provides the main development of skeletal muscle: as the body grows and strengthens, our muscles increase in size. After this period, the body begins to develop physiological decrements which will slow movements and reduce strength (Hayflick, 1980). As the body reaches its fifth to sixth decade of life, muscle size and strength tend to decrease significantly (Ballak et al., 2014; Proctor et al., 1998; Kyle et al., 2001; Mitchell et al., 2012; Lexell et al., 1988;

Metter et al., 1997; Lindle et al., 1997; Larsson et al., 1979) or adapt to provide relevant stability to particular areas of the body (Sierra et al., 2013). Some literature suggests that this decline can begin even earlier, around the third or fourth decades (Metter et al., 1997; Asmussen, 1980). This reduction in mass is termed Sarcopenia (Clark & Manini, 2008), or muscle atrophy, and can severely reduce the body's functional ability. The following sections give a brief overview of the changes in muscle mass, fibre type, strength and contraction velocity documented for an elderly MSK system.

2.4.2 Muscle Mass

There is a general agreement that muscle mass is lost with age. Studies have shown some loss of muscle mass can occur as early as the third decade (Janssen et al., 2000) but it is only around the fifth to sixth decades that significant reductions occur (Lexell et al., 1988; Janssen et al., 2000; Abe et al., 2014; Keller & Engelhardt, 2013). Narici & Maffulli (2010) found that from the second to eighth decade of a person's life, the skeletal muscle mass reduced by approximately 18% in men and 27% in women, up to extremes of 40% for those around the age of 80 (Proctor et al., 1998). It is also worth noting that loss of muscle mass does not appear to be accompanied by a loss in muscle strength (Maclennan et al., 1980; McPhee et al., 2013; Mitchell et al., 2012; Morse et al., 2005). This suggests that muscle architecture is not the only driving force behind strength.

The differences between the muscle architecture in males and females has been extensively explored and there is evidence to show that men and women atrophy at different rates (Gallagher et al., 1997; Abe et al., 2014; Maden-Wilkinson et al., 2013; Hughes et al., 2001). Gallagher et al. (1997) showed that despite having more skeletal mass than women, the rate of atrophy with age was greater in men. This was consistent across multiple muscle groups. This makes grouping genders difficult, and studies which require considerations of muscular strength should not ignore the significant differences between men and women, as well as the age-related changes.

2.4.3 Muscle fibre type

Decreases in the size and number of muscle fibres is a prominent contributor to both the loss of muscle mass (Narici & Maffulli, 2010) and function (Aoyagi & Shephard, 1992; Larsson et al., 1979). As discussed in Section 2.1.2.1, muscle fibre determines the strength and speed of a contraction. This reduction in the number of fibres, and any changes in type, will undoubtedly influence the ability for a muscle to perform the same function it was capable of at a younger age. In a review of Sarcopenia prevalence with age, Narici & Maffulli (2010) found numerous studies whose proposed that Type 2 (fast-twitch) fibres were on average 42% smaller (ranging between 20 and 50% - Ballak et al., 2014) in individuals aged between 55 – 60, relative to those in their twenties. Comparatively, Type 1 (slow-twitch) muscle fibres were only 23% smaller. Uncertainty between studies is to be expected, as the distribution of fibre types can vary across the body (Johnson et al., 1973; Lexell et al., 1983). If the body has severe Type 2 fibre atrophy, but Type 1 fibres are largely unaffected (which has been shown to occur - Lexell et al., 1988), then postural muscles (Type 1 dominant) will become more important whereas the ability to produce fast, strong contractions as a reaction to sudden external stressor events will diminish. There is evidence to support a shift towards a 'slower phenotype' with age in order to preserve vital movement functions (Sierra et al., 2013). This could explain the shift towards a higher ratio of Type 1 muscle fibres, as their role in postural, aerobic, muscle contraction is vital in reducing movement energy costs.

2.4.4 Muscle strength

Changes in muscle mass and fibre type provide evidence as to why muscle strength reduces with age, and it is universally agreed that the elderly are physically weaker. Reviews by Doherty (2003) and Aoyagi & Shephard (1992) put typical loss of strength into old age between 20 and 40%, and the following review will detail this for the upper and lower bodies. Table 2.3 documents a sample of the studies reviewed by Doherty and Aoyagi, in addition to others in the literature, and demonstrates the diversity of reductions in strength with age. The studies included in this review were chosen based on the consistent methodology in determining strength (maximum voluntary contraction – MVC) age ranges used, and specification of muscle groups.

Table 2.3 - Studies evaluating differences in muscle strength between older and younger men. Exclusively male data, unless otherwise stated (W = Women included). Y = Young, MA = Middle-Aged, O = Old, OL = 'Oldest', FL = flexors, EX = extensors and PF = plantarflexors

Study	Age range (SD)	Muscles tested	Method used to determine strength	Strength reduction relative to young
Larsson & Karlsson (1978)	O:61.6 (0.6), MA:54.5 (0.6), Y:26.1 (0.8)	Legs	Exertion against bar-mounted force transducer	O: 24%, MA: 6%
Asmussen (1980)	O:75, MA:55, Y:30	Leg	MVC	O: 40%, MA: 15%
Young et al. (1985)	O:75, Y:25	Quadriceps	MVC against force transducer	39%
Klitgaard et al. (1990) - control subjects	O:69 (1.9), Y:28 (0.1)	Vastus lateralis and biceps brachii	MVC	Knee EX: 44% Elbow FL: 32%
Overend et al. (1992)	O:70.7(1.3), Y:24.5 (1.5)	Dominant (kicking) side knee EX and FL	MVC against dynamometer	FL: 22.3% EX: 23.6%
Metter et al. (1997)	O:74.8 (2.5), MA: 55.5 (3), Y:26.8 (2.5)	Whole upper limb (both arms)	MVC against dynamometer combined with grip strength	MA: 12%, O: 26%
Hurley et al. (1998)	O:72, MA: 56, Y: 23	Quadriceps (both legs)	MVC	O:32%, MA:15%
De Serres & Enoka (1998)	O:74 (6), Y:28 (5)	Left arm elbow FL	MVC against dynamometer	28% (W)
Ivey et al. (2000)	O:69 (3), Y:25 (3)	Quadriceps (both legs)	MVC	28% ('Before training')
Klein et al. (2001)	O: 81 (6), Y: 23 (3)	Left arm elbow FL and EX (non-dominant)	MVC against dynamometer	FL:28% EX: 33%
Klein et al. (2002)	OL:86 (4), O:77 (1.3) Y:23 (3)	Non-dominant left arm elbow FL and EX	MVC against dynamometer	FL & EX: O - 18% & 25%, FL & EX: OL 40% & 39%
Valour et al. (2003)	O:67.4 (4), Y: 22 (3)	Dominant arm elbow FL	MVC against dynamometer	20%
Toji & Kaneko (2007)	O:69.1 (3.7), MA: 50.9 (3.5), Y: 21.2 (1.3)	Right elbow FL	MVC against Wilkie's arm ergometer	MA: 13.1%, O: 24.7%.
Kubo et al. (2007)	O:70.4(4.8), Y:26.4(3.7)	Right knee EX and PF	MVC against dynamometer	Knee EX: 48%, PF: 45%
McPhee et al. (2013)	O:74.4(3.3), Y:23.4(2.9)	Quadriceps and grip.	MVC against dynamometer	35% Quad, 22% grip (W)
Maden-Wilkinson et al. (2015)	O:72.2(4.9), Y:22.8(3.3)	Knee EX	MVC against isometric chair	40%
Wu et al. (2016)	O:66.8 (3.4), Y:23.7 (4.2)	Right knee EX and FL	MVC against dynamometer	EX: 48%, FL: 53%

The method of exertion against a dynamometer is a well-established tool for determining joint torques during isometric and isokinetic contractions, and is regularly used in clinical settings. Despite the scope for both types of contraction, the studies chosen for inclusion only assessed maximal isometrics. Some studies have shown little-to-no strength reduction, or even slight increases, in elderly strength during eccentric contractions (Thelen, 2003; Mitchell et al., 2012; Porter et al., 1995). The measurement of isokinetic contractions does allow a broader perspective of muscle strength, however the number of variables affecting concentric or eccentric contraction (range of motion during test, speed of contraction) make it difficult to compare multiple groups accurately (Osternig, 1986). There are experimental limitations to determining muscle strength through the use of dynamometry. While this is a well-used method, it is restricted to certain planar movements. The studies reviewed mostly look at flexion and extension of the knee or elbow. This does not take into account shoulder abduction/adduction for instance, which is of importance when looking at activities of daily living (Chapter 3) that rely on this action.

For the upper limb (generally for muscles crossing the elbow joint), the reduction in strength in older subjects is between 18 and 40% although it should be noted the upper end of this scale comes from one study which looked at both 'old' and 'oldest' age ranges (Klein et al., 2002). Excluding any data with age labels above 80, and mixed gender data, the mean value for upper limb strength reduction from men aged between 60 and 75 and those aged between 20 and 30 is 25 ± 5%. Some studies explored strength decline over multiple age groups, both between groups and longitudinally (Toji & Kaneko, 2007; Metter et al., 1997). However, the uncertainty around the start of strength decline with age means it is difficult to determine absolute strength reductions for a middle-aged group in a similar fashion to the above discussion.

A number of studies exploring changes of lower limb strength show that compared to the upper limb, the lower extremities weaken more rapidly with age (Lynch et al., 1999; Asmussen, 1980; Candow & Chilibeck, 2005) with a reduction in the range of 22 to 53% for the Quadriceps. This group of muscles feature predominantly in literature exploring lower limb strength (Ballak et al., 2014; Porter et al., 1995). Using the same age groups and inclusion constraints as for the upper limb literature data, the mean value for the lower limb strength reduction with age is 37 ± 9.7%. As the lower

limbs feature less frequently in ADLs used by the elderly, who mostly rely on the upper body, atrophy may be accelerated as a sedentary lifestyle does result in loss of muscle mass (Wisdom et al., 2014; Asmussen, 1980). The results of this are seen in a Health Survey for England by Whalley (2012), who reported that the elderly required more help for activities such as traversing stairs, taking a bath or shower, getting in and out of bed and generally getting around indoors (all predominantly lower limb activities) compared with washing, eating or toilet hygiene (primarily upper limb activities).

Generally, men are shown to be stronger than women at all ages (Ivey et al., 2000; Lindle et al., 1997; Frontera et al., 1991; Asmussen & Heebøll-Nielsen, 1962) but the differences are not always consistent as age progresses. Age reductions in strength are shown to decrease earlier and faster in men (Lindle et al., 1997; Lynch et al., 1999) although this does depend on the muscle group examined (Lynch et al., 1999; Asmussen & Heebøll-Nielsen, 1962). It would therefore be unwise to group men and women together when assessing changes in muscle strength with age. The review in Table 2.3 therefore only contains data for males.

2.4.5 Contraction Velocity

The ability of the body to produce strong contractions rapidly allows for safe and controlled reactions to external stressor events, such as falls or trips. Changes in fibre length, type and number (the latter two having been shown to decrease with age) will naturally change the muscle's contraction speeds (Narici & Maganaris, 2006; Thom et al., 2007). Thelen (2003) produced a set of literature-based values for a variety of muscle contractile properties when simulating dynamic muscle contractions in the elderly. They found large inconsistencies in quantifying a reduction in contraction velocity, but used the literature to produce a value of 20% loss of maximum muscle contraction velocity between young and elderly groups. Choosing values arbitrarily can reduce the validity of the muscle contraction computer simulations that Thelen was attempting to build (Hasson & Caldwell, 2012), however, literature data has repeatedly proven to be an inconsistent source of determining precise changes in the MSK system with age.

2.5 Concluding remarks

Muscle architecture changes dramatically with age, with a significant loss of muscle mass in the elderly. This is more prevalent in the lower half of the body. Contraction strength and velocity are reduced as a result, which appears to begin around the fifth decade of life. Due to differences in muscle mass and strength between men and women, grouping genders in MSK-related investigations would be ill-advised. It would be more sensible to carry out research on both groups separately, or focus on one gender with future work repeated for the other.

Comparing changes in strength between the young and the elderly, the latter have on average 25% less upper limb strength and a 37% reduction in lower limb strength. This difference between upper and lower extremity strength loss may change the movement strategy for activities requiring whole body motion.

As the body's strength reduces with age, its ability to perform an activity will decline. Many daily activities require a considerable amount of strength, such as ascending stairs, standing from a chair or carrying everyday items. Having explored the physiological changes in this chapter, the following chapter will discuss the consequences of a weakened body in relation to functionality. The condition of frailty will be introduced, a major consequence of a loss of muscle mass and strength, and the effects on performing activities pivotal to living a healthy life will be discussed.

CHAPTER 3

EXPLORING THE CONSEQUENCES OF AN AGEING POPULATION

The previous chapter discussed the changes within the body that occur with age, specifically in the musculoskeletal system. Chapter 3 introduces the concept of frailty, a condition affecting a large number of the elderly population. This chapter will cover basic definitions of frailty, and the standard measurement protocols in a clinical setting. This leads to the concept of musculoskeletal reserve (MSKR) and its relevance for ageing movement and physical demands on the body. The final sections will discuss the importance of activities of daily living for an ageing population.

3.1 Introduction to frailty

'Frailty' is the term used to describe a major consequence of the physiological changes explored in Chapter 2. Before any meaningful clinical interventions are implemented to combat the ageing body, the effects of frailty on everyday activities must be understood. Fried et al. (2004) gives the following definition:

"The physiologic state of increased vulnerability to stressors that results from decreased physiologic reserves, even dysregulation of multiple physiological systems"

During normal functioning our bodies are a network of complex systems which interact to allow biological processes to respond to these 'stressor events'. Systems such as circulation, the digestive tract or the neural network respond to perturbations by creating short, closed-loop circuits in order to restore the body to normality (Lipsitz, 2002). These bodily functions progressively decline with age. However, it is generally believed that it is the *number* of systems in decline, rather than the quantifiable deterioration of each system, that is important in defining frailty (Clegg et al., 2013). It would therefore stand to reason that a failure in the muscle remodelling process, and the systems which maintain healthy muscular tissue, can lead to a significant decline in the body's strength and mobility.

When a person is considered at the peak of their physical fitness, typically from the end of puberty through to early 30s, and they experience a physical stressor event such as a joint dislocation, bone breakage or muscle tear, they are able to recover within a specified time and continue to live an independent life. Those in their later years will experience these events and can expect an extended recovery period (Fried et al., 2001). If someone is considered severely frail, then that recovery period can become indefinite (Rockwood & Mitnitski, 2007). Those who reach a stable condition may never fully recover and would require around-the-clock care, becoming dependent on the healthcare system or family members.

There is a strong relationship between frailty and age in both men and women (Song et al., 2010; Fried et al., 2001; Rockwood et al., 2004), with prevalence in women higher than men (Collard et al., 2012; Fried et al., 2001; Rockwood et al., 2004) in terms

of both frailty and pre-frailty (Collard et al., 2012). Collard and colleagues acknowledge that this could come as a result of higher inclusion of women in the studies they reviewed. The fact that, on average, women have a longer life expectancy could also affect the data included in these studies. Despite this, the differences observed in ageing and frailty patterns between men and women make it inadvisable to group genders when investigating physiological parameters of any age group. This should be taken into consideration when forming studies based on functional ability or age.

The global prevalence of frailty is hard to determine due to the variance in assessment tools available, and between countries (Milte & Crotty, 2014; Santos-Eggimann et al., 2009; Collard et al., 2012). Studies suggest that over the age of 80, frailty affects between 15 and 50% of the elderly population (Clegg et al., 2013; Snih et al., 2009; Collard et al., 2012), and between 5 and 25% of people from the age of 65 to 80 (Milte & Crotty, 2014; Santos-Eggimann et al., 2009; Snih et al., 2009). Prevalence of pre-frailty, where an individual's health is less at risk than those classed as severely frail but still exhibits the signs of frailty, ranges from 35 to 46% in the over 65s (Romero-Ortuno et al., 2010; Snih et al., 2009; Collard et al., 2012). This shows that over half of those classified as elderly (over 65) can be considered as frail individuals, or at least exhibiting signs of developing frailty. There is also evidence of frailty and pre-frailty in the middle aged population (Santos-Eggimann et al., 2009), suggesting that the failing of internal systems at a younger age could be a pre-requisite for frailty later in life. With such significant numbers of the public classed as frail, and figures which estimate those of a frail disposition being more likely to be institutionalised (Clegg et al., 2013), it comes as no surprise that interest in understanding the effects of frailty have risen. Frailty in the elderly is therefore an issue identified as one of the main risks of an ageing population.

Despite the global nature of such a condition, there is little understanding of how frailty affects the physical aspects of everyday life. Muscle wastage, also termed sarcopenia, is a consequence of failing internal systems (Gielen et al., 2012) and will undoubtedly have severe effects on someone's functional abilities. Considering the physiological changes discussed in Chapter 2, and in tandem with the description of frailty, the strong connection between the two concepts shows a clear path for functional degradation. What follows is a downward spiral, where frail individuals will become

more sedentary, leading to further muscle atrophy. This is shown in Figure 3.1 where reduced physiological reserve and cellular damage can lead to a state of frailty, which in turn can lead to an increase in falls and disability when an external stressor (highlighted in red) is introduced. The dashed arrow represents the completion of the cycle, leading to further reductions in physiological reserves.

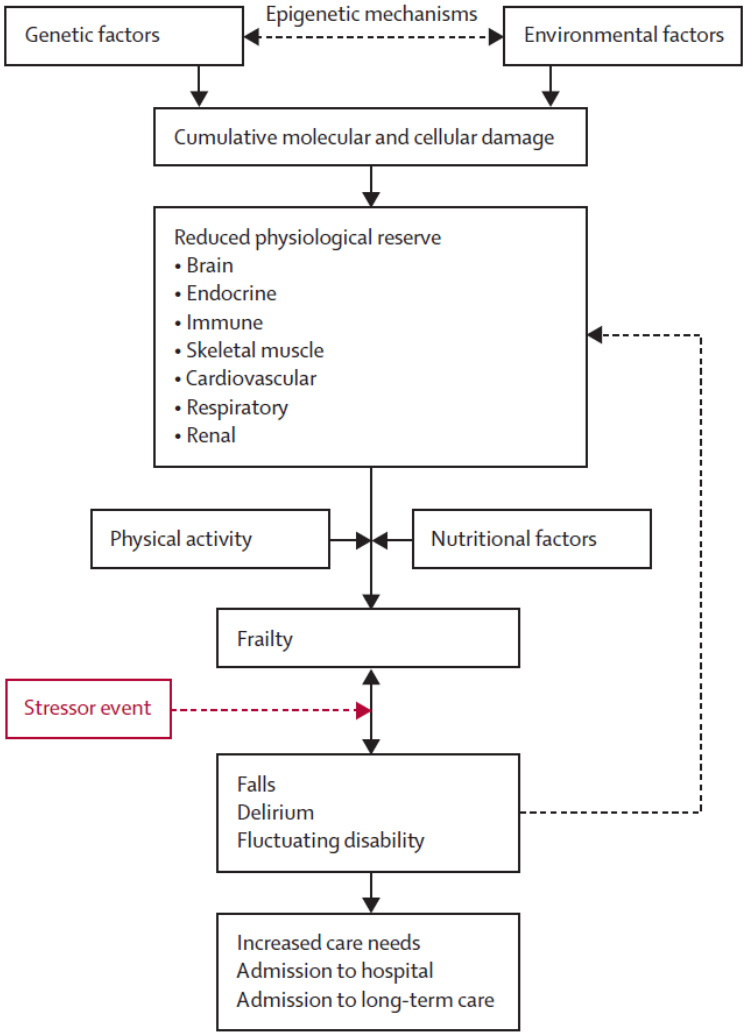


Figure 3.1 - Cycle of pathophysiology of frailty. Reproduced with permission from Clegg et al. (2013)

Past research has demonstrated strong links between physical frailty and disability (Fried et al., 2001; 2004; Boyd et al., 2005), but fails to fully quantify the effects of frailty on the body. Research instead makes links between frailty and muscle wastage (Fairchild et al., 2014; Cruz-Jentoft et al., 2010; Morley, 2008) or activities of

daily living (ADL) and instrumental activities of daily living (IADL) disability (Gobbens & van Assen, 2014; Gielen et al., 2012). Extensive reviews showed that training interventions, albeit when performed continuously, can play a positive role in rehabilitation (Theou et al., 2011; Liu & Latham, 2009). The review paper by Theou and colleagues showed that different age groups, levels of frailty and comparing men and women had different outcomes from these exercise regimes on physical determinants, functional ability and adverse health consequences. Evidently there is no single training programme suitable for frail elderly populations, with some programmes showing less than 30% positive outcomes (Theou et al., 2011).

Generalised training programmes, or simply knowing frailty and disability are linked, is of little use when trying to design focused rehabilitation or surgical interventions. The effects on movement can only truly be understood once we have a good understanding of the biomechanical effects on the body when it is weakening and less able to react to external stressor events. Before these can be quantified, the level of frailty must be assessed. Multiple studies have investigated this and produced tools for instant diagnosis, which is further discussed in the next section.

An alternative measure of ageing and functional decline is the later-life functional and disability instrument (FDI - Jette et al., 2002; Haley et al., 2002). The later-life FDI does not consider the elements of frailty, but instead allows the measurement of function in relation to disability in an older population. All measures of the functional part of this metric allow the assessment of function without assistance, whereas many other functional measures include questions based on completion with assistance. This questionnaire divides functional ability into 3 domains: advanced lower extremity functioning, basic lower extremity functioning and upper extremity functioning (Haley et al., 2002). This allows increased adaptability to individual activities, with an indication as to which are more important to functioning for each patient. However, the choice of activities, for example unscrewing a lid, picking up a kitchen chair and running half a mile appear to be unnecessarily specific and may not be applicable to a general population. These functional based measured are valid for studies not considering all elements of frailty, however, such work would not consider the psychological and social aspects of functional decline. Therefore, a more comprehensive measure of frailty in all domains would be required.

3.1.1 Assessment of frailty

The highest level of frailty assessment is a comprehensive geriatric assessment (CGA). A CGA is used as a multidimensional diagnostic tool to determine a frail individual's general health, mental health and functional capacity (Martin, 2010), and will determine how at-risk an elderly person may be prior to hospitalisation or further treatment. This level of assessment has been shown to reverse the progression of frailty, improve functional ability and increase independence in the elderly population (Stuck et al., 1993; Ellis & Langhorne, 2004).

Numerous groups have developed tools and questionnaires in an attempt to diagnose frailty for use in clinical settings, where individuals can be identified as being at risk due to worse outcomes amongst the frail compared to healthy elderly (Wyrko, 2015). It is important that such tools are easy to use, comprehensively cover all aspects of frailty and provide the same level of assessment as a geriatrician, regardless of whether a specialist is administering the test or not. Wyrko (2015) identified two main frailty models, from which most tools are developed: the frailty phenotype model (Fried et al., 2001) and the cumulative deficit model (Mitnitski et al., 2001; Rockwood & Mitnitski, 2007). To briefly summarise, the phenotype model consists of five variables determined from a prospective group not initially designed to investigate frailty (Clegg et al., 2013): weight loss, self-reported exhaustion, low energy expenditure, slow gait speed and weak grip strength; where the presence of 3 or more (frail), 1 or 2 (pre-frail) or none (not frail) determined frailty status. This is evidently simple to administer, however, it may miss important factors such as cognitive impairment (Clegg et al., 2013). The cumulative deficit model is similar to the phenotype, but considers a much larger group of variables; up to 92 'deficits' in total (Wyrko, 2015). Similar to the phenotype model, the more deficits present in an individual, the more likely they could be classified as frail (Clegg et al., 2013). Due to the large number of variables, this model provides a more accurate grade of frailty which would allow a clinical setting to quantify patient requirements. However, there is a much larger time commitment to complete the assessment and it is best administered by someone with extensive clinical experience.

Choosing the correct frailty tool is important to optimise patient care and identification of those at greater risk. de Vries et al. (2011) compiled a comprehensive

list of assessment tools developed over the last 30 years, and rated the ability of each to cover a range of eight clinical factors. They found that no single test was suitable for all clinical settings, and that the focus of the individual test determined its suitability. For example, a researcher with little clinical experience may find a self-reported test more useful, whereas a performance based test would be of more use to a trained clinician. A successful tool would cover all three domains of frailty assessment: physical, psychological and social, and de Vries and colleagues found that few covered all three. In particular, social support was poorly assessed, with only a handful of the tools including related questions (de Vries et al., 2011). Overall, the tool developed by Rockwood and Mitnitski (Mitnitski et al., 2001) was determined to cover all frailty factors. However, for research similar to that presented in this thesis, a screening tool with a far simpler set of questions is required.

Over half of the recognised frailty factors are centred around the physical aspects of ageing (strength, physical activity, mobility, nutritional status - de Vries et al., 2011). The importance of physical frailty, in relation to the musculoskeletal changes explored in Chapter 2, highlights the need to focus on the physical side of frailty. It is also necessary that the tool chosen covers some or all of the above factors.

One such tool is the Edmonton Frail Scale (EFS, Rolfson et al., 2006), which was not included in the work carried out by de Vries and colleagues as their review did not take screening tools into account. The EFS has been identified as an easy-to-use, broad assessment of frailty, recommended for use by the British Geriatric Society and suitable for both specialists and novices to administer (Wyrko, 2015; Partridge et al., 2012; Turner & Clegg, 2014). In terms of validation, the EFS was compared with the results of a CGA and administered by both a research assistant with no formal medical training and a geriatric specialist. The results showed good correlation with the CGA and had good inter-assessor consistency. There are only 11 questions to administer and it can be performed independently or with the help of a carer, despite lack of clinical experience. Table 3.1 presents the test in full, and shows how the EFS covers all three frailty assessment domains as identified by de Vries et al. (2011): physical – through a functional independence question and a timed-up-and-go test, psychological – a cognitive test and mood questions, and social – a question on available support.

Having explored both the definition and assessment of frailty, and the deficit in research to fully quantify its effects on the body, it is necessary to develop a system by which the physical aspects of a frail individual can be measured. The decline of the MSK system in terms of mass, volume and contractile ability is well documented, and the continual decline in functional ability has been assessed through both the research and clinical settings. However, the question remains: how precisely does a lack of strength affect our ability to respond physically to a stressor event? If the effects of frailty are to be understood on a physiological scale, this phenomenon must, and can be, quantified. The following section will explore this further.

Table 3.1 - The Edmonton Frail Scale. A score of 0-5 = Not frail, 6-7 = vulnerable, 8-9 = mild frailty, 10-11 = moderate frailty and 12-17 severe frailty (Rolfson et al., 2006)

Domain	Item	0 points	1 point	2 points
Cognition	Clock test – Please imagine that this pre-drawn circle is a clock. I would like you to place the numbers in the correct positions then place the hands to indicate a time of 'ten after eleven'	No errors	Minor spacing errors	Other errors
General health	In the past year, how many times have you been admitted to hospital?	0	1-2	≥ 2
	In general, how would you describe your health?	'Excellent' 'Very Good' or 'Good'	'Fair'	'Poor'
Functional independence	With how many of the following activities do you require help? Meal preparation, shopping, transportation, telephone, housekeeping, laundry, managing money, taking medicine.	0-1	2-4	5-8
Social Support	When you need help, can you count on someone who is willing and able to meet your needs?	Always	Some-times	Never
Medication use	Do you use five or more different prescription medications on a regular basis?	No	Yes	
	At times, do you forget to take your prescription medications?	No	Yes	
Nutrition	Have you recently lost weight such that your clothing has become looser?	No	Yes	
Mood	Do you often feel sad or depressed?	No	Yes	
Continence	Do you have a problem with losing control of urine when you don't want to?	No	Yes	
Functional performance	Timed up and go test – Start with arms and back at rest. Stand and move to 3 m mark and return	0 – 10 s	11 – 20 s	One of: >20 s, patient unwilling or requires assistance
Totals	Final score is the sum of all questions		/ 17	

3.2 Musculoskeletal reserve

In order to understand how physical frailty may affect an individual's ability to complete activities of daily living (ADLs), a quantifiable measure must be developed. As discussed previously, there are numerous tools by which frailty can be assessed but the collateral effects on functional ability of such a condition have not yet been investigated. 'Musculoskeletal Reserve' (MSKR) was a term introduced by Bull et al. (2008) to define the additional biological reserves each individual has to complete a task, in this case the reference was to physical reserves of muscular strength. It was surmised that this reserve determines whether or not the body can overcome an external stressor such as injury or sickness in order to continue to function independently. Failure to do so could lead to further health problems, dependence on others, institutionalisation or severe frailty. It is already known that this reserve is in-built into the human body, as it possess more muscle mass as well as neuronal circuitry, renal nephrons and hormonal stores than are required for our survival (Lipsitz, 2002). It can therefore be initially hypothesised, due to the earlier descriptions of frailty, that this reserve of biological systems is reduced in the elderly. It is this anomaly of strength deficit that requires further investigation.

3.2.1 Exoskeleton design – 'Assistance as needed'

A good example to further explain the concept of MSKR is in the design of exoskeletons. Exoskeletons are generally non-invasive, orthotic devices which assist the user through mechanically driven motion. An example of such a device is shown in Figure 3.2. As with any orthotic, prosthetic or joint replacement, it is important that the human body is as active as possible without heavy reliance on the external device. For example, stress shielding is a common consequence of using implants with material properties stiffer than that of bone (Huiskes et al., 1992; Bobyn et al., 1992). This leads to weakening of the cortical bone as the foreign material is able to take the strains usually placed on the biological structures. The use of exoskeletons must follow the same precautions, as it is all too easy for a user to fully rely on the mechanical device to perform everyday movements. Carmichael & Liu (2013) investigated the concept of 'Assistance-as-needed', or AAN, as a paradigm for controlling assistive robotics with

the underlying message that it is vital for an individual to be as active as possible during physical rehabilitation (Talkowski et al., 2009; Afschrift et al., 2014). The AAN concept bridges the gap between the body's ability to perform a desired task and the task's physical requirement for completion. A visual representation of this is shown in Figure 3.3.



Figure 3.2 - Example exoskeleton product which assists the patient in moving their lower extremities (Medical Expo, 2017)

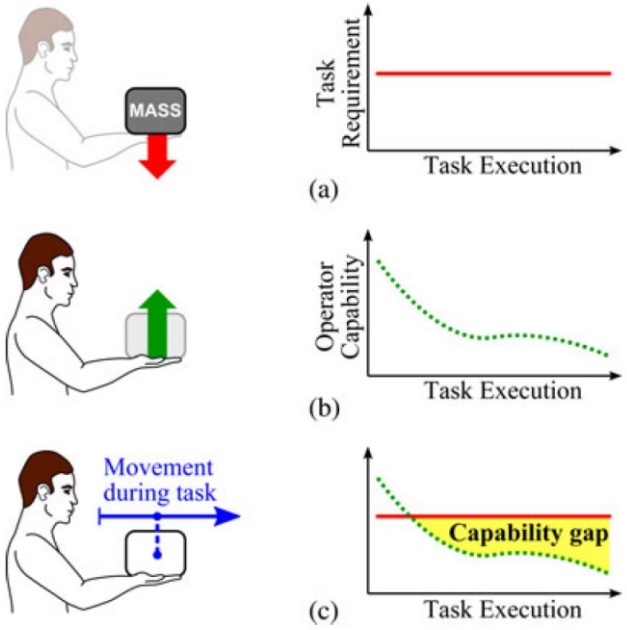


Figure 3.3 - Visualisation of the assistance as needed concept: (a) shows the strength requirement in order to perform the task effectively, (b) shows the predicted capability of an individual over time and (c) highlights the gap between these two (Carmichael & Liu, 2013). Reproduced with permission.

It is worth noting that the predictions of the user's strength in this study were shown to be lower than those previously published in the literature. This begs the question as to how the AAN will respond, most likely providing too much assistance when the user is able to complete the task independently. This also highlights one major limitation of quantifying MSKR: how can maximum strength be measured? Neither AAN nor MSKR can be applied clinically if there is no measurement of the patient's original strength. The use of dynamometry is a popular solution to this problem (Holzbaur et al., 2007; Stark et al., 2011). However, this does not provide maximum strengths for individual muscles but rather functional groups. There remains the need to predict a user's strength profile based on known physiological facts about the functional abilities of older generations.

Functional demand (a task's physical requirement) has previously been investigated (Samuel et al., 2013; Hortobagyi et al., 2003; Hughes et al., 1996) and shown to increase dramatically in the elderly during tasks such as ascending or

descending stairs and sit-to-stand, leaving minimal extra capacity if the body experiences an unsuspected fall or further injury. Considering the changes that are known to occur in ageing musculature (Chapter 2), it would stand to reason that from a physiological standpoint MSKR would diminish over time. Hortobagyi et al. (2003) showed that task completion for an elderly group required between 78 and 88% of their maximum strength levels for basic activities, compared with 42 to 54% in the young. This will undoubtedly produce a natural prejudice against performing tasks independently, instead relying on external devices such as walking sticks, supportive frames or the previously given example of exoskeletons. In order to fully test and quantify the concept of MSKR, the movements and joint powers associated with ADLs should be explored. The next section looks into functional ability during these ADLs, and begins to focus in on those of most importance to an ageing population.

3.3 Activities of daily living

ADLs are an important assessment tool in clinical practice, and are seen as the actions most used throughout a typical day. A review by Murray produced a list of popular ADLs commonly investigated, and therefore seen as clinically relevant when determining functional ability. These activities included, but weren't limited to: personal hygiene (brushing teeth or washing the face), feeding oneself and more general actions such as standing and sitting, lifting everyday objects or driving (Murray, 1999). Subsequent studies made use of these activities to test early versions of musculoskeletal simulations (Charlton & Johnson, 2006; Murray & Johnson, 2004) and numerous others have since used similar activities as the basis for joint loading and functional ability studies (Anglin & Wyss, 2000; Anglin et al., 2000; Bergmann et al., 2007; Westerhoff et al., 2009; Hortobagyi et al., 2003; Gausden et al., 2017). Most of these ADLs are concerned with the action of the upper body, however, function of the lower body is also of importance in terms of general mobility and maintaining independence. There are no definitive lists as with the upper limb, however common tests for functionality with a heavy focus on the lower limb include timed-up-and-go tests (Donoghue et al., 2014; Beauchet et al., 2010; van Iersel et al., 2008), gait analysis (Potter et al., 1995; Donoghue et al., 2014; van Iersel et al., 2008) and the ability to traverse stairs (Bean et al., 2007).

As the body ages, it becomes increasingly difficult to perform these ADLs (Hortobagyi et al., 2003; Samuel et al., 2013; Landers et al., 2001; Covinsky et al., 2003) which can lead to both physiological and psychological health risks (Markides et al., 1996; Mori et al., 2017; Williamson & Fried, 1996). A health survey for England found that anywhere up to 30% of over 65s could not perform simple tasks such as traversing stairs, shopping for essentials or doing housework (Whalley, 2012). In a study of nursing home residents, it was found that between 68 and 80% of individuals required assistance in performing ADLs such as going to the toilet, walking, eating and general mobility (Mehr et al., 1993). These are large portions of the elderly population who are unable to take care of themselves, and combined with increased healthcare costs in treating those of a frail disposition (Comans et al., 2016; Snih et al., 2009; Fried et al., 2004), it becomes vitally important to understand the physiological consequences of functional decline. This will reduce the strain on the healthcare system and increase independence in an ageing community.

The ability to perform some of the more important ADLs has been observed in order to understand how an ageing body contributes to functional degradation. Ascending and descending stairs is an extremely strenuous task for the elderly (Startzell et al., 2000) and can require joint moments exceeding those available to an ageing body (Hortobagyi et al., 2003). Carrying heavy loads independently can be tiring (Ayis et al., 2007) and recovery may vary depending on the weight carried and distance travelled. One particular activity, identified as one of the more difficult to achieve at old age and requires whole body movements, is standing from a seated position (Alexander et al., 1991; Wheeler et al., 1985; Van der Heijden et al., 2009; Dehail et al., 2007; Hughes & Schenkman, 1996). Inability to complete a sit-to-stand (STS) task can lead to an increase in care (Perry et al., 2006; Fisher et al., 2010) and a higher risk of further problems in other ADLs (Gill et al., 1995), compared to those of a similar age who are more independent. To rise successfully out of a chair, the body needs to reach a stable standing position which the elderly may struggle to do unassisted. Between 8 and 31% of non-institutionalised over 65s have difficulty getting in and out of beds and chairs (Dawson et al., 1987; Gill et al., 1995).

Rising from a chair is a demanding task, requiring high amounts of joint torque and core strength (Schultz et al., 1992; Hughes et al., 1996; Yoshioka et al., 2012). It

has been previously shown that lower-body joint power is the defining characteristic of STS performance (Corrigan & Bohannon, 2001; Lord et al., 2002; Rantanen et al., 1994). Hughes et al. (1996) found that up to 97% of available strength was being used by an elderly group in order to achieve successful chair rise. Energy costs of performing functional activities are significantly higher in the elderly (Layec et al., 2014; Malatesta et al., 2003; Mian et al., 2006), and considering the susceptibility they have to fatigue (Leng et al., 2002; Artz, 2008), it stands to reason that the elderly will be working relatively harder to complete a STS motion compared to the young. In addition to the physiological issues associated with performing such a high demand activity, there are psychological problems relating to reported pain and anxiety (Lord et al., 2002). A recent study correlating functional disability to 'a sense of life worth living' showed that poor self-esteem can result in a rapid decline in activity (Mori et al., 2017). These factors are difficult to quantify, and therefore this thesis will focus on the biomechanics of STS. However, these previous studies also do highlight the need to better understand the non-physiological factors in STS performance, and that an inability to complete ADLs is multi-factorial.

Therefore, it could be hypothesised that the ability to perform a STS motion does not mean continued success into later life. Difficulties in chair rise can be attributed to muscle weakness, and as a consequence the elderly have to change strategy in order to achieve the motion (Dolecka et al., 2015; Van der Heijden et al., 2009). However, studies which have previously investigated the STS movement are limited in that they only measure functional muscle group power and cannot fully quantify the contributions of individual muscles. Due to the synergistic and antagonistic activity of all muscles in joint movement, it is difficult to definitively say which muscles should be targeted for rehabilitation through physiotherapy or simple strength training. Further work is needed in order to focus the clinical application of improving STS performance.

In order to complete the standing motion when suffering from muscle weakness, an individual will employ different strategies to make the best use of their remaining strength. Janssen et al. (2002) reviewed various studies which investigated determinants of the STS motion and found that speed, foot position, trunk position/movement and armrests were the most prolific strategy-related adaptations for all age groups. Of the 39 papers reviewed by Janssen, only 17 included subjects aged 65 and above. Concerning

the use of arm rests, due to the inherent importance for the elderly (Munro & Steele, 1998): only one study was found to specifically compare young and old groups with and without arm rests, with the remainder investigating changes in speed and chair height. There was, however, interesting work investigating increased trunk flexion (Alexander et al., 1991; Dolecka et al., 2015; Gross et al., 1998; Papa & Cappozzo, 2000) and feet movement (Dolecka et al., 2015; Papa & Cappozzo, 2000). The work by Papa & Cappozzo found that the elderly had an increased trunk flexion, at a higher velocity, after having left the seat compared to their younger counterparts. This also meant their centre of mass was closer to the support base (at the feet) which increased stability but required greater muscular effort to achieve. The study also found that elevation of the body only occurred once the centre of mass was in position over the feet, whereas the young would flex and rise synchronously. Postural stability, the success of an individual to maintain their centre of mass (CoM) over the base of support, at lift off appeared to be more important than required joint torques (Alexander et al., 1991; Schultz et al., 1992), suggesting the goal of STS is to achieve it safely rather than efficiently or comfortably. Akram & McIlroy (2011) firmly believed that to prevent balance issues, it was crucial to maintain CoM within the base of support. Therefore, leaning forward when not using armrests is a logical compensation, as those with lower stability will look to position their CoM over the feet or knee joints. From a seated position, this can require considerable strength which an older individual may not possess. Even for those achieving the task, the strain placed on the muscles and joints involved may cause severe discomfort and lasting damage. It has also been shown that hand use does not necessarily reduce required joint torques but, similar to increased trunk flexion, increases postural stability (Schultz et al., 1992).

Despite the hypotheses drawn up by the authors of the research reviewed above, and the evident importance to further investigate how reduced strength affects performance, there is a marked lack of research investigating the effects of using arm rests despite this being the simplest form of STS assistance. Those who have investigated the use of arm rests have shown a significant reduction in hip and knee moments when rising (Arborelius et al., 1992). However, there is very little extra work, and no work investigating loading of both the upper and lower limbs together. This highlights STS, in particular the use or lack of arm rests, as an interesting focal point on joint loading and effort required.

3.3.1 Phase Segmentation

The whole sit-to-stand-to-sit motion can be divided into two main portions: 'Stand' and 'Sit'. Previous studies have identified up to three phases into which the 'Stand' portion can be further subdivided (Millington et al., 1992; Duffell et al., 2013; Schultz et al., 1992; Schenkman et al., 1990; Houck et al., 2011). The first phase (Phase 1) consists of building momentum at the trunk, generally known as 'weight shift' (Millington et al., 1992) or 'flexion momentum' (Schenkman et al., 1990). The aim of this phase is to build enough momentum to lift off the seat with sufficient postural stability. Initiation of upper body movement is followed by a sharp increase in trunk flexion angular velocity (Schenkman et al., 1990). Phase 1 ends just prior to seat lift-off, however, Duffell and colleagues also presented an interim phase (termed Phase 1.5 for the sake of this thesis) between the point at which the seat force centre-of-pressure reaches its most anterior point and lift off. The justification for this small phase is unclear, and appears to simply separate the end of flexion momentum and leaving the seat. The duration is negligible, approximately one tenth of a second (Duffell et al., 2013), and any significant events could easily be placed into earlier or later phases.

The last phase (Phase 2) consists of the time just after seat lift off and ends when the subject reaches the final standing position. This can be identified by the hip joint angular velocity first reaching 0°/sec (Schenkman et al., 1990; Duffell et al., 2013), indicating the end of hip extension. Any variation in the angular velocity after this point can be attributed to swaying of the body when stood still.

3.4 Concluding remarks

This chapter has explored the consequences of an ageing population, and how the physiological changes reviewed in Chapter 2 can lead to conditions such as frailty. Those diagnosed as frail, or pre-frail, may experience extreme weakness, poor health and an increased need for care compared to those of a similar age but in better health. Various tools for identifying frailty were reviewed, and the EFS was found the most comprehensive, and suitable for use by a non-clinician with little geriatrics experience. Questions and test within this scale on the physical aspects of frailty make it suitable for

use in a MSK study, and as such will be administered to all future participants regardless of age. This ensures all control subjects meet a minimum functional level.

A frail individual may lead a more sedentary lifestyle due to these limitations in strength, or through psychological problems. This can continually lead to atrophy of muscle mass, and severe reductions in the available strength that an ageing body possesses. Differences in prevalence between men and women, in both ageing and frailty, mean separate groups should be considered for such research. The higher prevalence of frailty in women will require considerations for safety, particularly in the lab environment.

The concept of MSKR is introduced and will be used to understand the effects of physical frailty. Finally, the effects of age on functional ability were explored, specifically highlighting the importance of rising from a chair and how the high demands of the task can lead to further problems in the elderly body. It is clear from this review that linking the idea of a reduced strength capacity with the inability to stand up and sit down may lead to a better understanding of why the elderly struggle with such activities. As rising from a chair is a highly demanding task, one which in later years the majority of the elderly simply cannot perform, it would be suitable for assessing the effects of strength reduction and how reduced MSKR could be a cause for difficulties.

CHAPTER 4

MUSCULOSKELETAL MODELLING OF THE UPPER AND LOWER LIMBS

This chapter will give a brief overview of the upper and lower limb musculoskeletal models used in this thesis.

4.1 Introduction

Musculoskeletal (MSK) models are tools used to analyse the mechanics of movement, providing information on physiological parameters difficult to measure in vivo, such as muscle activations and joint contact forces. There are two approaches to the biomechanical analysis of an MSK system: Inverse Dynamics (ID) and Forward Dynamics (FD). The latter require large amounts of computational power as these predict motion based on joint torque or muscle excitation data. This requires implementation of complex muscle contraction mechanics, for example in the form of a Hill-type muscle model (Hill, 1938). This model element requires knowledge of individual muscle force-length-velocity relationships (introduced in Chapter 2) and tendon mechanics, allowing the conversion of muscle excitation to contraction. Unless the model is highly constrained, this becomes a vastly complicated problem (Otten, 2003). Alternatively, ID models (those used within this thesis) use observed kinematics as the primary input and, as a result, require less computational power to quantify the muscle forces required for each motion. A number of MSK models, utilising either ID or FD, have been previously developed to analyse movement. These include, but are not limited to: FreeBody (Cleather & Bull, 2015; 2011; 2010; Cleather et al., 2011; Ding et al., 2016), the UKNSM (Charlton & Johnson, 2006), the DELFT elbow and shoulder model (Nikooyan et al., 2011), the Stanford-VA model (Holzbaur et al., 2005) and the platforms of OpenSim (Delp et al., 2007) and AnyBody (Lemieux et al., 2012). AnyBody is a commercial package used to create bespoke MSK models but there are few publications and as such is not a complete, valid model. Both OpenSim and AnyBody have a partial open-source approach (OpenSim more so than AnyBody), and are therefore open to alterations depending on the research. However, there are aspects of the models that are not able to be amended easily, these include the ability to have a floating centre of rotation of the knee joint (a key factor in quantifying forces at the knee), or the ability to use direct measures of scapular tracking. The Stanford-VA model, unlike the DELFT and UKNSM models, lacks comprehensive musculature at the shoulder, and would not provide the information required for this thesis. By comparison, the UKNSM features 87 lines of action at the shoulder and elbow and includes the option of scapula-specific tracking for a full understanding of bespoke shoulder kinematics. FreeBody allows for the use of a floating centre of rotation of the knee and provides a more accurate assessment of knee

joint forces (Ding et al., 2016). Therefore, the UKNSM and FreeBody are used in this thesis.

This chapter will briefly outline the processing steps behind ID modelling, and introduce the two MSK models used within this thesis.

4.2 Musculoskeletal modelling

Measured movements act as the primary inputs to ID models, predominantly obtained using optical motion capture. A flowchart of the MSK modelling steps which quantify the muscle and joint forces associated with this motion data is shown in Figure 4.1. Anatomical datasets, such as those compiled by Klein Horsman et al. (2007) and the Visible Human Project (Spitzer & Whitlock, 1998), provide bone geometries and muscle architecture for the initial scaling sections of MSK modelling. With body segment measurements, the information provided by these datasets can be scaled to a subject for bespoke muscle insertion and origin points, for example. The accuracy of MSK modelling is thought to increase when a range of anatomical models are considered (Ding et al., 2016). In these cases, the height and weight of the current subject are approximately matched to the height and weight of the subjects used to build each dataset.

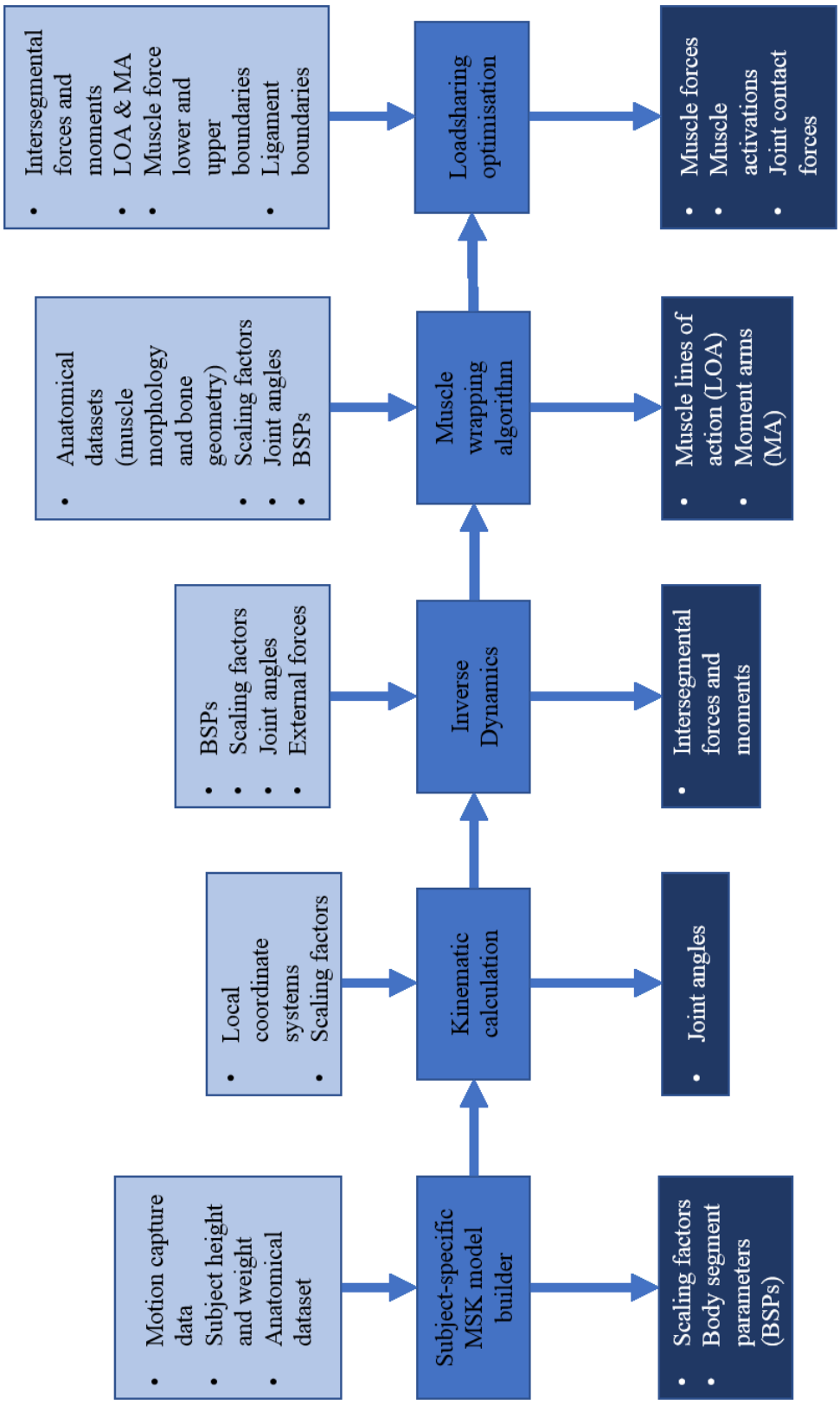
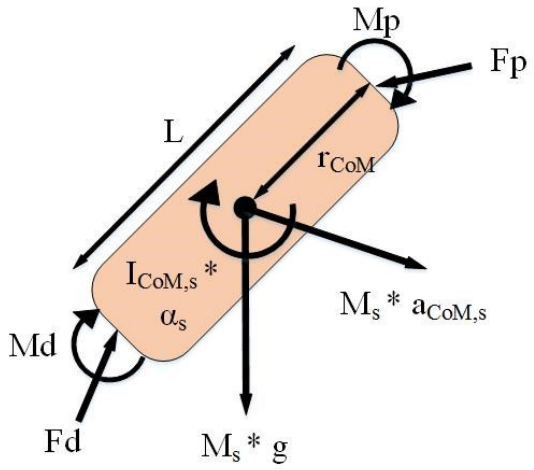


Figure 4.1 - Flowchart describing the steps of ID MSK modelling. The middle row states each modelling step, the required inputs are shown in the top row and the outputs are shown in the bottom row

Body segment parameters (BSPs) are also scaled based on regression tables, such as those developed by De Leva (1996), using a subject's height and weight. Segment mass, centre of mass (CoM) and inertial properties are important in calculating intersegmental forces and moments (Figure 4.2). The accuracy of the scaled model is of particular importance in the context of an ageing body, and when comparing genders. Research has shown the distribution of segment mass to differ with age and gender (Jensen & Fletcher, 1994) and basing multiple subjects on one anatomical dataset may introduce errors. This is discussed further in Chapter 5, where a subject-specific method of obtaining BSPs is presented.



$F_{p,d}$ = Joint force at the proximal/distal end

$M_{p,d}$ = Intersegmental moment at the proximal/ distal end

M_s = Segment mass

L = Segment length

r_{CoM} = Distance of Segment centre of mass (CoM) from proximal end

$I_{CoM,s}$ = Segment moment of inertia about the CoM

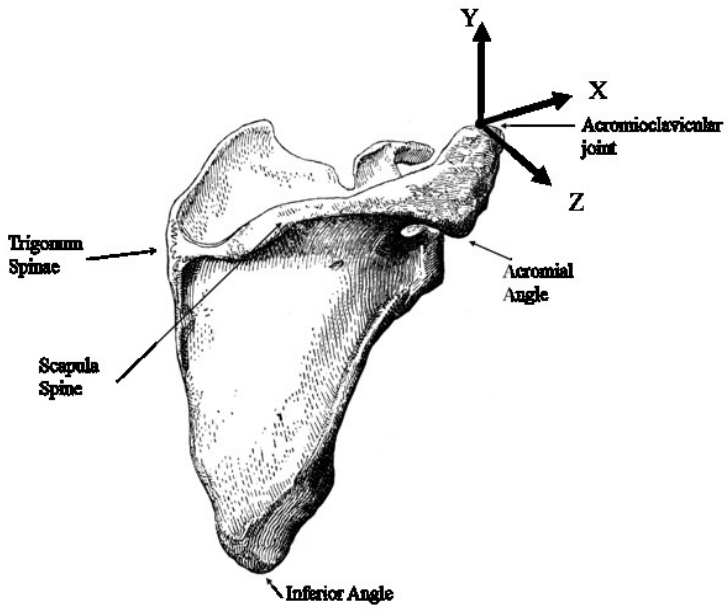
α = angular acceleration

$a_{CoM,s}$ = linear acceleration of the CoM

Figure 4.2 – 2D, generic-segment, free body diagram. BSPs used within MSK models have been highlighted in bold on the right-hand list

Once the scaled anatomical model for each subject has been created, segment coordinate frames can be found. These are used to locate each body segment in space, and are calculated based on anatomical landmarks found manually prior to data collection during motion capture. The conventions for these frames are standardised (Wu et al., 2002; 2005), and an example of how to construct the anatomical frame of the scapula is given in Figure 4.3. Detailed information on the local coordinate frames used within the United Kingdom National Shoulder model (UKNSM) and FreeBody

can be found in Prinold's thesis (2013) and Appendix 1 respectively. The joint angles are calculated by extracting the Euler angles of each axis from the transformation matrix between two local coordinate frames.



$$\bar{X} = \frac{\overline{AA} - \overline{TS}}{|\overline{AA} - \overline{TS}|}$$

$$\bar{Z} = \frac{\bar{X} \times (\overline{AI} - \overline{TS})}{|\bar{X} \times (\overline{AI} - \overline{TS})|}$$

$$\bar{Y} = \bar{Z} \times \bar{X}$$

Figure 4.3 - Scapula local coordinate frame and equations for calculating each axis. AA = acrmomial angle, TS = trigonum spinae and AI = inferior angle. The origin of the scapula frame lies at the acromioclavicular joint (Blaisdell, 1904)

From this, the intersegmental forces and moments are calculated using the Newtonian equations of motion. A simplification of these equations is shown below (using the abbreviations from Figure 4.2):

$$Fp = M_s.a_{CoM,s} - (Fd + M_s.g) \qquad \text{Equation 4.1}$$

$$Mp = \left(I_{CoM,s}.\alpha_s + r_{CoM}.M_s.a_{CoM,s}\right) + (Md + L.Fd) + (r_{CoM}.M_s.g) \qquad \text{Equation 4.2}$$

The muscle paths can be constructed once the joint angles are calculated and the MSK model is scaled accordingly. Muscle forces act along their muscle line of action, formed between the insertion and origins points previously scaled to the subject. Computer simulations therefore model the muscle paths as straight lines

(essentially taut strings). Many muscles are positioned around bones which are represented in MSK modelling by wrapping objects, also scaled to an individual. Within the model the position of these wrapping objects, and the subsequent changing muscle paths, are calculated for each time frame of the observed motion. For every time step, the path the muscle takes around the object is calculated as the shortest distance between the insertion and origin points. However, at extreme joint ranges of motion, this can result in 'flipping' of the path (Figure 4.4) through a non-physiological direction of the wrapping object (Persad, 2016). This is of particular concern when calculating shoulder wrapping objects, due to the large range of motion of this joint relative to less complex articulations of the body. The use of via points (Persad, 2016) and energy minimisation techniques (Marsden et al., 2008) have been introduced to resolve wrapping issues at extreme ranges of motion.

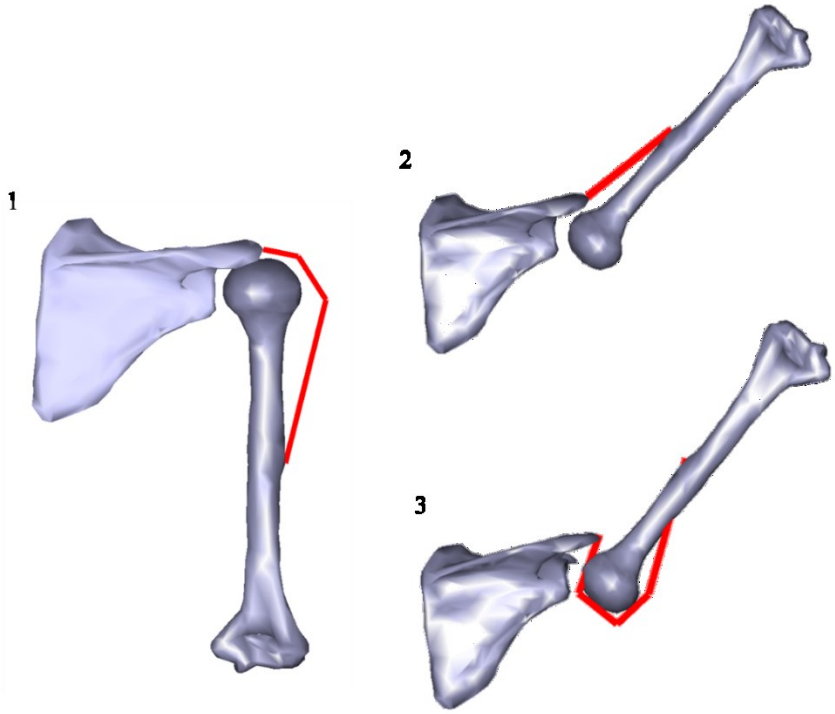


Figure 4.4 - Example of muscle flipping of the deltoid muscle. 1 - At rest, where red line represents deltoid line of action, 2 - Humerus abducted by 150°, with the red line following the correct path of the deltoid after contraction and 3 - muscle has 'flipped' to a non-physiological line of action due to an error in the muscle wrapping algorithm

Once calculated, and shown to be physiologically appropriate, muscle lines of action are used to determine the acting moment arms of each muscle. Combining the known intersegmental moments and individual muscle moment arms allows the calculation of muscle forces, where the moments exerted by the muscles are mathematically equal to the intersegmental moments. Due to the static indeterminacy of the human body, the number of muscles which could be driving the motion is larger than the number of equations available. Therefore, mathematical optimisation is used as a load-sharing solution. This optimisation solves the set of equilibrium equations by quantifying muscles forces within set constraints and through the minimisation of a cost function. The functions used in this study are described in Section 4.3 and Section 4.4.

The mathematical constraints usually consist of upper and lower boundaries for the muscle forces and ligaments modelled. Muscles force upper boundaries (MFUB) are calculated based on physiological cross-sectional area (PCSA), which differs from anatomical cross-sectional area as it describes the packing of muscle fibres as opposed to actual size, and specific tension (or maximum muscle stress):

$$MFUB = PCSA * specific\ tension \qquad \text{Equation 4.3}$$

The MFUBs determine the boundaries of the solution space for the optimisation routine. Again, Sections 4.3 and 4.4 introduce additional boundaries that can be included in individual models. The resultant muscle forces can then be used to calculate components of the joint reaction forces. The following sections will introduce each MSK model featured in this thesis.

4.3 The United Kingdom National Shoulder Model

The first of the two models is the United Kingdom National Shoulder Model (UKNSM - Charlton & Johnson, 2006). The UKNSM is predominantly concerned with the structures in and around the shoulder and elbow (Figure 4.5), including four joints, three ligaments, 87 muscle elements (grouped into 29 functional parts forming 21 muscles) and all upper limb bones (except the hand), all of which were introduced

in Chapter 2. Bone geometry for anatomical scaling is taken from the Visible Human Dataset (Spitzer & Whitlock, 1998) and muscle morphology from the cadaveric work of van der Helm et al. (1992), Johnson et al. (1996) and Veeger et al. (1997).

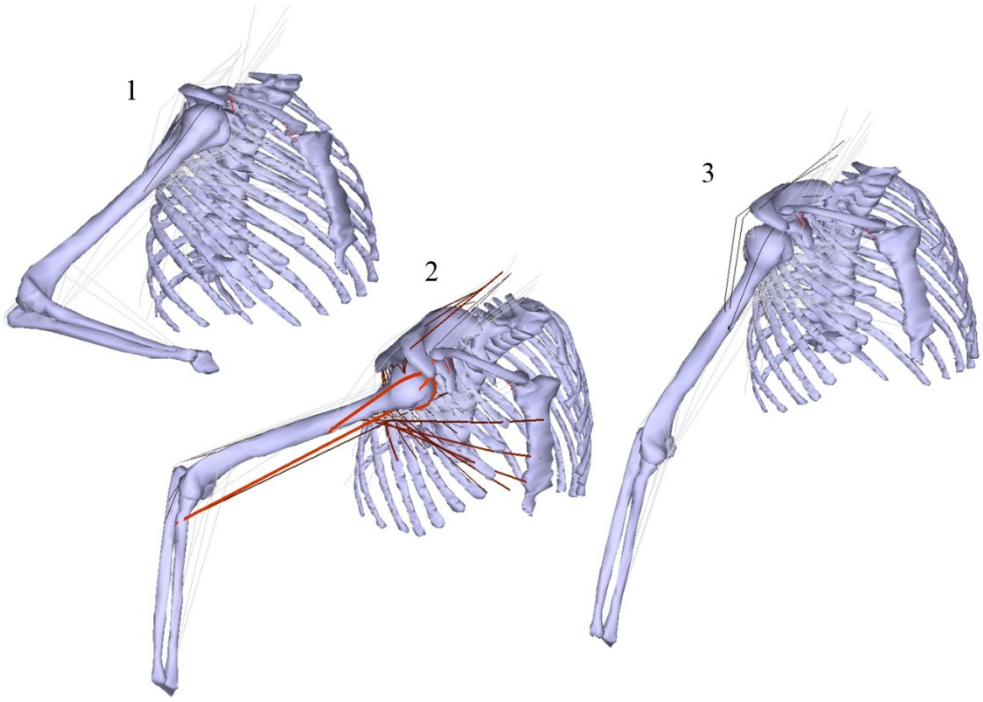


Figure 4.5 - UKNSM visualisation of a STS trial. 1 – Sat using a chair armrest, 2 – mid-rise, 3 – standing position. Black lines represent muscle lines of action, a change to red indicates activation and a deeper shade of red shows increases in activation

The protocols governing the running order of the UKNSM follow the basic rules set out in Section 4.2, with the addition of a kinematic optimisation. This optimisation determines the position of the scapula, as tracked during the motion capture stage and determined by calibrations described in Chapter 7. Typically, the scapula kinematics are calculated via regression and depend on Scapulohumeral rhythm (Barnett et al., 1999). However, the work of Prinold et al. (2011) and Shaheen et al. (2011) has provided the means to track scapula movements during motion capture data collection. Both Prinold (2012) and Persad (2016) provide detailed explanations of how the optimisation constrains the scapula position based on anatomical landmarks.

After publication in 2006, numerous researchers have improved the functionality of the UKNSM: introducing improved muscle wrapping protocols

(Marsden et al., 2008; Persad, 2016), bespoke scapula tracking (Prinold et al., 2011; Shaheen et al., 2011) and further improvements to the kinematic optimisation to constrain the scapula based on tracked, anatomical landmarks during extreme shoulder ranges of motion (Persad, 2016). The cost function of the UKNSM minimises the sum of squared muscle stresses:

$$\sum_{i=1}^{87} \left(\frac{F_i}{PCSA_i}\right)^2 \qquad \text{Equation 4.4}$$

$$and\ LB < F_i < UB$$

Where i = muscle element, F_i = estimated muscle element force of each iteration, $PCSA_i$ = the PCSA for each muscle element and *LB* and *UB* are the muscle element lower (this is typically set to 0 N) and upper force boundaries respectively. Scaling of the muscle force upper boundaries is not as straight forward as other physiological parameters, and is further discussed in Chapter 7. In addition to these boundaries, the UKNSM also constrains the point at which the humeral head contact force is applied to the glenoid cavity. This is achieved through the representation of the glenoid as an ellipsoid, and by constraining the joint force vector to lie within the 2D ellipse. This simulates joint stability at the shoulder.

The outputs of the UKNSM are validated through comparisons with alternative MSK models (Charlton & Johnson, 2006) which includes models with direct comparison to instrumented implant data (Nikooyan et al., 2010 - Comparison performed by Prinold, 2012); a general review of moment arm calculation (Gatti et al., 2007); and partially through electromyography (Johnson & Pandyan, 2005; Pandis et al., 2015). Published work using the UKNSM includes analysing joint contact and muscle forces during driving (Pandis et al., 2015), extreme motions such as pull-ups (Prinold, 2012), cricket bowling (Persad, 2016) and labour-intensive motions such as brick-laying (Ismail, 2016).

4.4 Freebody

The second model used in this thesis is the open source software FreeBody (Figure 4.6 - Cleather & Bull, 2015; 2011; 2010; Cleather et al., 2011; Ding et al., 2016).



Figure 4.6 - Freebody visualisation of a gait trial. Red lines represent muscle lines of action, where a deeper colour means higher activation. Arrows represent joint contact forces

In the same way as the UKNSM, FreeBody utilises ID for its biomechanical analysis, and features muscles and bones in and around the hip, knee (both tibiofemoral and patellofemoral joints) and ankle. This includes positioning of the patella bone (Cleather & Bull, 2015), which is important when considering loading of the knee joint (Masouros et al., 2010).

FreeBody features 163 muscle elements, grouped into 57 muscles (see Chapter 2 for details), and 14 ligaments. Muscle morphology is taken from the anatomical dataset compiled by Klein Horsman et al. (2007) and, more recently, a range of new datasets for improved customisation (Ding et al., 2016). The load-sharing optimisation for FreeBody is also constrained by the upper and lower boundaries of muscle forces and uses a cost function based on muscle activation:

$$\sum_{i=1}^{163}\left(\frac{F_i}{F_{max,i}}\right)^3 \qquad \text{Equation 4.5}$$

$$and\ LB < F_i < UB$$

Where $F_{max,i}$ = muscle activation as determined by the ratio of F_i to UB. In a similar fashion to the UKNSM restraining the shoulder contact force, FreeBody can constrain the rotational joint centre of the hip to reduce the sensitivity of the segment-based approach to skin motion artefacts (Ding et al., 2016), a feature used in this thesis. The study by Ding and colleagues, primarily investigating the effects of subject-specific anatomical datasets, also serves as a validation of both quantified contact forces and muscle activations via comparison with instrumented implant and electromyography (EMG) data. Previous work using FreeBody includes the simulation of functional electrical stimulation (Rane & Bull, 2016), forces during jumping and heavy lifting (Cleather, 2010) and work on improving patient specific contact forces at the knee (Ding et al., 2016).

4.5 Concluding remarks

The models presented in this chapter are both validated and suitable for use in this thesis. Outputs of muscles forces, and calculated upper boundaries, allow the assessment of muscle activation (muscle force relative to maximum force) for quantification of musculoskeletal reserve, and calculated joint contact forces will help better understand the forces on the body during the sit-to-stand movement. Intersegmental moments (joint torques required for the observed movement) may explain the enhanced requirements for motion completion discussed in Chapter 3. Despite the accuracies of the models presented, there are still differences when calculating the scaling factors between anatomical datasets and subject-specific physiological data. Due to known alterations in segment mass distributions with age, it is necessary to explore the effects of such inconsistencies and develop methods by which BSPs can be determined on a bespoke level. The following chapter will discuss the potential consequences of using regression tables to determine BSPs, and present an alternative method for bespoke measurement of these parameters.

CHAPTER 5

MEASURING SUBJECT-SPECIFIC BODY SEGMENT PARAMETERS

This Chapter investigates the use of 3D scanning to calculate bespoke body segment parameters which act as subject-specific inputs into MSK models, as described in Chapter 4. The 3D scanning method is compared with cadaveric data of segment mass, centre of mass and inertial properties, as well as against geometric modelling for subject-specific validation. Group-specific regression tables are explored as an alternative to large population-derived tables.

5.1 Introduction

Musculoskeletal modelling requires body segment parameter (BSP) inputs, such as segment mass, centre of mass (CoM) and inertial properties, in order to quantify intersegmental muscle and joint forces. These are typically calculated through regression-based anthropometric tables, such as those developed by De Leva (1996), and require inputs of the subject's height and weight. As these tables are usually based on small populations, the reliability in predicting an individual's BSPs is subject to the anthropometrics of the cadavers/living subjects used to calculate the regression coefficients. Pearsall & Costigan (1999) estimated the BSPs for the shank and thigh multiple times using regression tables from five published regression tables, and found the mass and inertial properties to vary by up to 40%. Their research compared regression tables populated with cadavers and living subjects, males and females, and age groups ranging from 15 to 83 years. This seems illogical, considering changes in BSPs have been shown to be dependent on both age and gender (Durkin & Dowling, 2003; Pavol et al., 2002; Jensen & Fletcher, 1994). Durkin & Dowling (2003) also compared literature regression tables and found that differences in predicted segment masses and centre of mass measurements varied between 30 to 150% and 60 to 250% respectively depending on segment, gender and age group. Errors between measured and regression-predicted BSPs are shown within Durkin & Dowling's own work; producing differences of up to 18% when comparing the dual-energy x-ray absorptiometry (DEXA) measurements of segment mass against the same study's calculated regression data.

Anthropometrics have also been shown to vary by race. Shan & Bohn (2003) showed differences in leg length between German and Chinese men and women, which, in terms of gait patterns, may not have significant effects on stride length (Cavanagh & Kram, 1989; Steudel-Numbers et al., 2007) but can change the energy requirements of walking (Steudel-Numbers et al., 2007). Steudel-Numbers et al. (2007) showed that the energy cost of locomotion increased with mass, therefore, the joint torques required to overcome associated limb inertia and facilitate movement are greater. If regression tables are predicting erroneous segment masses, this will affect the validity of MSK modelling outputs. This is particularly true for non-homogenous body types, or those whose mass distribution across segments differs from those used in published regression

tables. Research into the changing characteristics of ageing muscle shows that muscle mass (Narici & Maffulli, 2010; McPhee et al., 2013) and body segment volume (Vidt et al., 2012) is reduced for both men and women in the later stages of life. In addition to this, the body tends to lose muscle mass in the lower limbs prior to losing strength in the upper part of the body, as discussed in Chapter 2.

There is conflicting evidence on the requirement for accurate quantification of BSPs. The sensitivity of MSK modelling outputs, such as joint moments and muscle forces, to variations in BSPs has been shown to be both significantly and insignificantly different (Rao et al., 2006; Dao et al., 2009; Wesseling et al., 2014; Fantozzi et al., 2005; Ganley & Powers, 2004; Myers et al., 2015; Sheets et al., 2010; Pearsall & Costigan, 1999; Nguyen & Reynolds, 2014; Lee et al., 2009). Wesseling et al. (2014) found that increasing shank CoM by 40% had the largest influence, specifically on hip flexion moments during gait. They also explored simultaneous perturbations of CoM and mass with the same outcomes, however the resulting changes in joint moments was reportedly small. The changes in BSPs employed by Wesseling do appear to be arbitrary (coincidently in line with the errors shown by Pearsall & Costigan, 1999), with no evidence to suggest ranges of error in line with previously reported regression tables, despite numerous studies available for review. Rao et al. (2006) also found that simultaneous changes in inertia, mass and CoM caused significant differences in hip joint moments, with deviations around 20% when comparing to data from unperturbed BSP inputs. This work, as with the majority of other studies investigating joint power sensitivity to BSP differences, based variation in BSPs on literature regression predictions. Other studies also identify hip joint moments as the most sensitive amongst lower limb joints to BSP variation (Fantozzi et al., 2005; Ganley & Powers, 2004; Sheets et al., 2010; Pearsall & Costigan, 1999; Myers et al., 2015). This was generally more evident during the swing phase of gait.

An increased sensitivity at the hip joint to BSP error could also be attributed to joint moment contributions in gait. During swing, where no additional external reaction forces are present, BSPs will dominate contributions to joint torque requirements, whereas ground reaction forces perform a similar role in stance. The majority of studies reviewed here used gait trials to determine the sensitivity of joint moments, however, some predict that for high acceleration activities the sensitivity may be higher (Prinold,

2012; Pearsall & Costigan, 1999). Many applications of musculoskeletal modelling are for low acceleration activities, where the inertial tensor falls out of the equations as angular accelerations drop to near-zero. Therefore, it is expected that low acceleration activities would be less sensitive to perturbations in BSPs (particularly inertia) than those with high accelerations. One study showed high sensitivity in joint moments, again mostly in the hip, during cycling (Wheat & Barratt, 2015), however, these uncertainties were small at 10% of the peak hip joint power. Wheat & Barratt predicted that the sensitivity to BSP variation would be of higher significance during competitive cycling as the rotations per minute (rpm) increased towards more realistic speeds. There are few studies looking at high-demand activities and the effects of BSP variation, and further research is needed to confirm these hypotheses. This falls outside the aims of this thesis, but developing a means by which this BSP sensitivity research can be achieved should be explored. Considering the potential effects on both quantified joint power and muscle force for certain activities, the aim should not be to improve upon the accuracy of existing methods but rather to develop protocols by which users can produce subject-specific BSPs rapidly. Bespoke methods of calculating BSPs should allow users to produce appropriately scaled models of individual subjects, as opposed to estimating these from regression tables produced from non-homogenous populations.

The development of group-specific regression tables, where the populations used to create the coefficients have a defining feature, such as: gender, being within a specific age range or BMI groupings, may be one way to improve BSP specificity in the absence of more bespoke methods. As previously discussed, there are two significant constraints when using literature regression tables:

1. Small numbers of subjects/cadavers used to populate the tables and;
2. The anthropometry and details of the subjects/cadavers used will most likely not match a new study's group of interest.

Previous literature suggests there are differences in BSP absolute values and relative proportions between different ages (Durkin & Dowling, 2003; Muri et al., 2008; Chambers et al., 2010) genders (Jensen & Fletcher, 1994; Durkin & Dowling, 2003; Chambers et al., 2010; De Leva, 1996; Pavol et al., 2002; Rossi et al., 2013) and

morphologies (Chambers et al., 2010; Damavandi et al., 2009). Therefore, using regression tables would be ill-advised for specific groups which have large variations. Constructing these tables is a relatively simple task, however, the time gathering data will determine if the regression tables are necessary or not. Sukits et al. (2011) found that comparing the regression model of De Leva (1996) with models built based on age, gender, height and BMI showed significant differences for old, obese adults. The size of all groups used did not exceed 13 subjects (Chambers et al., 2010), and the age range was limited (65-85 years), however, this exhibited the potential improvements in BSP estimation via group-specific regression.

This chapter will review previously used methods for calculating subject-specific BSPs, including volumetric, surface and geometric methods. A study will then be presented looking at the use of infra-red depth sensors to produce BSPs with quick-360-degree scans, the results of which will be compared to cadaveric studies and geometric modelling. Finally, use of group-specific regression models will be discussed to determine whether the presented protocols could be used as a rapid method of calculating regression BSP tables for specific populations such as the elderly groups that form a key part of this thesis.

5.2 Review of methods used to obtain BSPs

There are three main methods for determining BSPs: regression, geometric estimation and 3D surface scanning. Regression involves building BSPs with values relative to height and weight and based on small populations from direct measurement (Section 5.1). As these are usually smaller studies, other methods such as gamma-ray scanning (Zatsiorsky et al., 1990) or cadaver-based experimental methods can be used to build the initial database. As reviewed above, the errors associated with regression equations can be large. However, they are quick and easy to use for studies where the movements are slow or consider large additional external forces (excluding gravity) on the body. These render segment parameters negligible when applied to the body, such as holding objects or using support during ambulation. In these cases, the variation would not be significant enough to affect MSK modelling outputs.

Studies which utilised direct measurement techniques are presented in Table 5.1, which includes a mixture of cadaver and living-based techniques. All cadaver based studies used some, or all, of the following direct measurement methods: weighing, use of a balance-board, water immersion and pendulum-based, to determine weight, centre of mass, volume and inertial properties respectively. Each of these studies produced their own regression coefficients and tables, based on the unique population used.

Geometric modelling is an alternative method, whereby instead of using population-based regression, the user takes a number of anthropometric measurements and approximates the shape of body segments using known geometric forms. This can be time-consuming in a clinical setting. Hanavan (1964) developed a mathematical model of the body using geometric shapes for calculating inertial properties (Table 5.2). The model requires 25 anthropometric dimensions in order to calculate centres of mass and moments of inertia. Each shape gives an estimate of segment volume that, when combined with calculations of uniform density, give segment mass. However, it has been shown to be less accurate than alternative models in predicting these properties (Durkin & Dowling, 2003). The errors are attributed to inaccuracy in standard shapes representing large, complex body segments and through the assumption of a uniform density value within segments; this is known not to be the case for all body tissue types (Winter, 2009). This is unavoidable in many cases due to the difficulty in obtaining bespoke density profiles, and previous attempts have struggled to prove the improved accuracy of such expensive methods (Wei & Jensen, 1995).

Table 5.1 - Studies which measured/estimated body segment parameters for mass, centre of mass and/or moments of inertia in order to produce regression tables. Gender symbols are used to identify males (♂) and females (♀)

Study	Tissue type	Parameter collection/estimation method	Subject information (ages given in years)
Harless (1860)	Cadaver	-	Mass/CoM: 2 ♂, aged 29 & one unspecified Volumes: 7 ♂ and ♀, aged 20 – 68.
Braune & Fischer (1889)	Cadaver	-	4 ♂, aged 18, 45, 50 and one unspecified
Dempster (1955)	Cadaver	-	8 ♂, aged 52 - 83
Drillis & Contini (1966)	Living	Immersion, reaction changes and compound pendulum methods	20 ♂, aged 20-40 years
Clauser et al. (1969)	Cadaver	-	13 ♂, aged 28 – 74
Chandler et al. (1975)	Cadaver	-	6 ♂, aged 45 - 65
Clarys & Marfell-Jones (1986)	Embalmed Cadaver	-	3 ♂ (aged 15, 69,78) and 3 ♀ (aged 79,79,81)
Zatsiorsky & Seluyanov (1990)	Living	Gamma-ray scanner	100 ♂ (mean age 24) and 15 ♀ (mean age 19)
Jensen & Fletcher (1994)	Living	Two cameras and digitising software	7 ♂, 12 ♀. Aged 63 to 75 years
De Leva (1996)	Living	Adjustments to Zatsiorsky & Seluyanov (1990), to change the mean relative centre of mass positions and radii of gyration to be referenced by joint centres	See Zatsiorsky & Seluyanov (1990)
Durkin & Dowling (2003)	Living	Whole body dual energy x-ray absorptiometry (DEXA)	4 age groups of 25 subjects: young/old, ♂ /♀, aged 19-30 and 55+

Table 5.2 - Shapes and measurements used to define body segments for geometric modelling (Hanavan, 1964)

Segment	Shape used to represent segment	Anthropometric measurements required
Head	Circular ellipsoid	Head height and circumference
Upper torso	Elliptical cylinder	Chest breadth, waist depth and distance from shoulder acromion to substernal (Xiphoid process)
Lower torso	Elliptical cylinder	Waist depth, buttock depth and distance from substernal to top of the buttocks
Hand	Sphere	Fist circumference
Upper arm	Frustum of circular cone	Upper arm circumference, elbow circumference and upper arm length
Forearm	Frustum of circular cone	Elbow circumference, wrist circumference and forearm length
Upper leg	Frustum of circular cone	Thigh circumference, knee circumference and distance from bottom of the buttocks to the knee
Lower leg	Frustum of circular cone	Knee circumference, ankle circumference and distance between knee and inferior portion of malleolus
Foot	Frustum of circular cone	Distance from ground to inferior portion of malleolus and foot length

Three-dimensional imaging techniques have been used to obtain BSPs quickly, without the need for numerous anthropometric measurements and avoiding potentially inaccurate regression tables. Techniques such as magnetic resonance imaging (Martin et al., 1989; Cheng et al., 2000; Mungiole & Martin, 1990) and DEXA (Durkin & Dowling, 2003; Ganley & Powers, 2004; Wicke & Dumas, 2014; Lee et al., 2009) give good estimates of BSPs but are limited by cost, extensive post-processing, significant assumptions related to tissue density, and, in the case of DEXA, exposure to ionising radiation.

More recently, 3D light scanning has been introduced as a method to obtain rapid measurements of the body without expensive equipment or time-consuming scan times.

These can either be through the use of laser (Sheets et al., 2010; Lu & Wang, 2008; Pandis & Bull, 2017) or infra-red light (Norton et al., 2002), neither of which pose any harm to the body, unlike some of the aforementioned medical imaging techniques. The use of infra-red scanners, such as the Microsoft Kinect (Microsoft Corporation, Redmond, USA), has proven popular not just in the gaming industry (its original purpose) but in robotics, body scanning and healthcare environments (Clarkson et al., 2015). Its low cost, retailing at £80 (Microsoft, 2017), readily available software for integration with motion capture environments and promising results in estimating BSPs (Kudzia, 2015) and object scanning (Menna et al., 2011) make the Kinect a good alternative for full body scanning.

Despite the numerous advantages, many full body scanners are large and cumbersome, featuring multiple sensors and frames requiring assembly. One particular scanner, produced by Styku (Los Angeles, USA), has been designed to minimise these disadvantages and would give a motion capture environment quick access to 3D anthropometric data of the body. A full body scan with this product takes less than a minute and produces a surface-point cloud of between 40,000 and 60,000 vertices. The scan is produced by an Xbox Kinect which remains stationary whilst the subject is rotated about a single point on a small platform (Figure 5.1). The platform rotates at approximately 6° per second and performs one-full 360° rotation. The space required is small: only 2.5 x 1.7 m is sufficient for a full scan, however more space may be needed depending on arm span.

The aim of the following sections will be to assess the use of a 3D infra-red positional scanner, the *Styku 3D full body scanner*, in producing subject-specific BSPs for the potential use as bespoke inputs into MSK models.

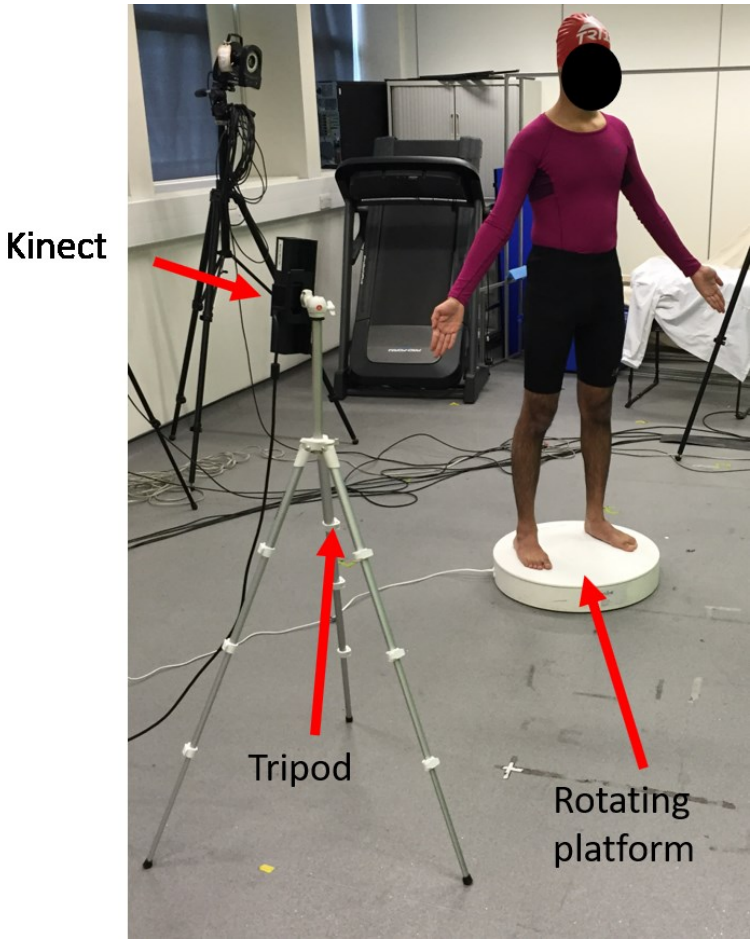


Figure 5.1 - Full body 3D scanner setup with a participant in the anatomical stance

5.3 Methods

This study was approved by the Joint Research Compliance Office and Imperial College Research Ethics Committee. The full application and approval letter can be found in Appendix 2.

95 males and females volunteered for a full body scan using the Styku 3D body scanner. Each participant was given the opportunity to read the study information before giving written informed consent. Part of the consent detailed that the data obtained from the scan would be entered into an anonymised database whereby future studies could make use of the data. One participant asked not to be included in this database.

Each subject completed a 21 question survey detailing basic information on age, sex, height (measured using a stadiometer – Seca 213, Seca, Germany), weight (measured using standard scales – 9048 BK3A, Salter, UK) and hand dominance. It also included questions on profession, working hours, children, exercise habits, diet, general health and if they suffered from a list of conditions including arthritis, diabetes, stress or bulimia or had any history of joint replacement surgery (Appendix 2). This information could be used in future studies looking at correlating subject-specific BSPs and lifestyle. Additionally, this could be used to identify participants for group-specific regression. The subjects then changed into skin tight clothing (a form-fitting running kit was provided), removed their shoes and socks and either tied their hair or placed it under a swimming cap. Before starting the scan the equipment was calibrated, which involved removing the subject and any objects (other than a partial view of the rotating platform) out of the scanner's field-of-view. This allowed the scanner to identify the platform and set the global origin at the platform's centre. The subjects then stepped onto the rotating platform with their feet shoulder-width apart, to assume the anatomical stance as shown in Figure 5.1. The scan was initiated and the platform operated remotely. After approximately 60 seconds, the platform was halted and the scan stopped.

All data were output as vertices points and face designation. These detail the coordinate of each vertex and row indication for triangulation purposes when reconstructing the 3D image. The body was segmented as described in Section 5.3.1 and BSPs were calculated for each segment as shown in Figure 5.2. This process is described in more detail throughout Section 5.3.2.

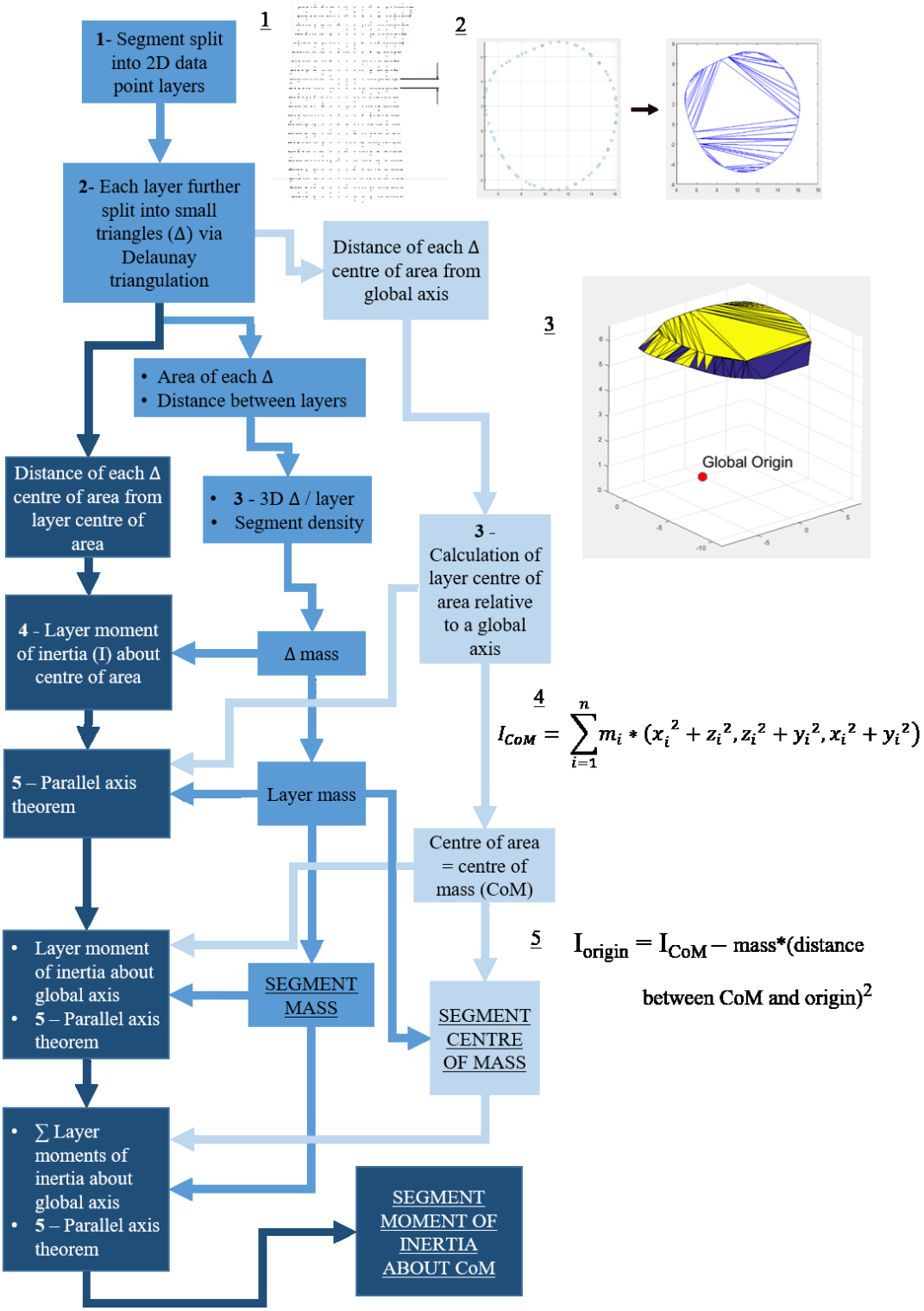


Figure 5.2 - Flowchart describing the steps for calculating the body segment parameters for individual segments. Dark Blue = steps for inertial property calculation, Blue = mass, and Light Blue = centre of mass. Numbers in boxes correspond to images/equations

5.3.1 Scan segmentation

Anatomical landmarks were manually identified for splitting the scans into body segments (Table 5.3), where each segment was based on anatomical articulation (Harless, 1860). The thigh segment consists of the mass from the knee joint up to a plane with passes through the groin and hip joint centre (De Leva, 1996) which would follow the lateral edge of the pelvis as defined by the anterior superior iliac spine (ASIS), face of the acetabulum and ischiopubic ramus (Drake et al., 2010). Due to a lack of visual external landmarks on the scan to define the right and left ASIS points, the distance from knee joint to hip joint centre was estimated based on the expected thigh length (as per De Leva, 1996) and the shank ratio of measured to expected length. Some studies draw a cutting plane parallel to the transverse plane through the groin (Davidson et al., 2008; Giachetti et al., 2015; Zhiqiang et al., 2012) which would not accurately represent the thigh as a functional segment as the most superior point of the groin does not align with the hip joint centre.

The skull was separated from the neck in order to leave the head as a single segment. The plane for cutting the head and neck followed the lateral line of the inferior mandible to the inferior portion of the occipital bone (Drake et al., 2010). The remaining portion of neck was discarded as it does not form a functional segment. The upper arm included data for the shoulder, and so the proximal boundary followed a plane joining the armpit and acromion points. The feet, shanks, forearms and hands were all segmented based on planes through anatomical joints; namely the ankle, knee, elbow and wrist.

Table 5.3 - All anatomical landmarks and planes used to partition body into functional segments. Any bony landmarks stated were visually identifiable on the 3D scans but this list also allows for palpation in a lab setting if reproduced. Planes are shown in Figure 5.3

Segment	First plane	Second plane
Foot	Plane connecting lateral/medial malleolus	-
Shank	Plane connecting lateral/medial malleolus	Horizontal plane passing through the most anterior point of the patella
Thigh	Horizontal plane passing through the most anterior point of the patella	Plane originating at the most medial part of the groin and passing through the estimated hip joint centre
Trunk	Plane originating at the most medial part of the groin and passing through the estimated hip joint centre	Plane connecting the jugular notch and the most posterior-inferior part of the neck (or the seventh cervical vertebrae)
Upper arm	Plane connecting the armpit and most superior-lateral point of the shoulder	Plane perpendicular to the arm long axis, passing through the olecranon process
Forearm	Plane perpendicular to the arm long axis, passing through the olecranon process	Plane connecting the wrist styloids
Hand	Plane connecting the wrist styloids	-
Neck	Plane connecting the jugular notch and the most posterior-inferior part of the neck (or the seventh cervical vertebrae)	Plane connecting the most posterior-inferior point of the occipital bone and a point where the mandible meets the neck line
Head	Plane connecting the most posterior-inferior point of the occipital bone and a point where the mandible meets the neck line	-

5.3.2 Body segment parameter calculation

Once the body was partitioned into 14 individual segments (Figure 5.3), each was passed through an algorithm calculating bespoke BSPs, as described in Figure 5.2. As the scans give volumetric profiles of each segment, estimations of density were required to calculate segment mass.

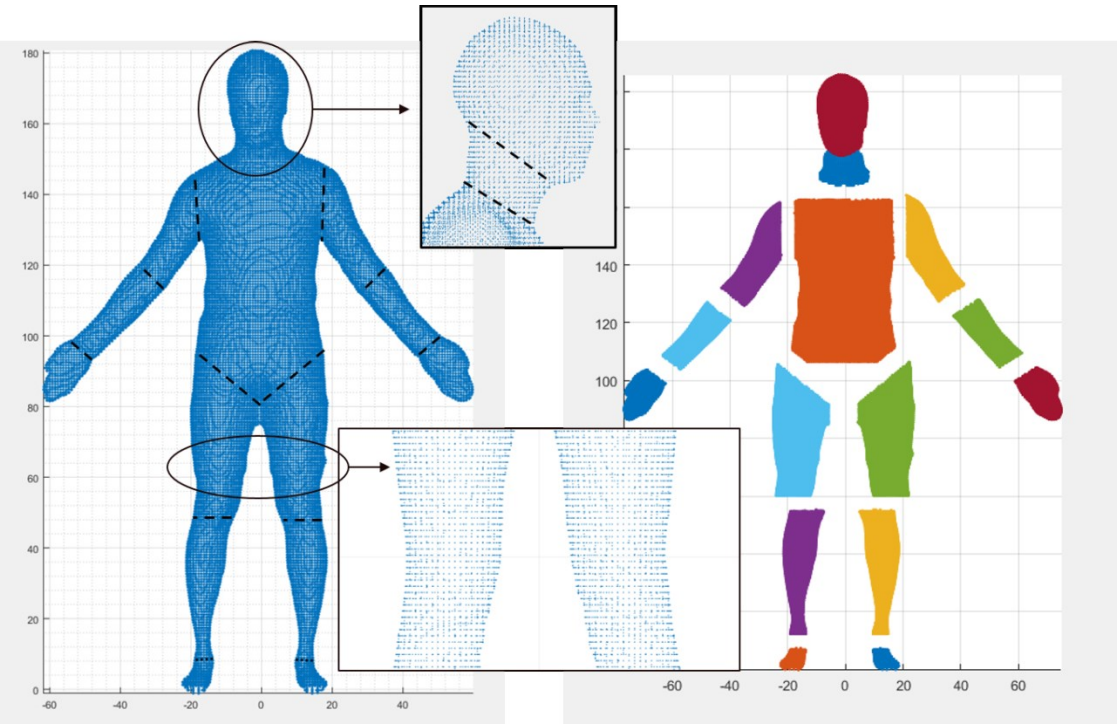


Figure 5.3 - Initial data (left) with segmentation planes indicated by dotted lines (top boxed image shows planes for the head and neck), zoomed in section of thighs to show layers parallel to transverse plane (centre) and segmented scan (right)

5.3.2.1 Density

Due to the difficulties in obtaining bespoke density profiles for individual segments and tissues within segments, it is common to assume a uniform density within a segment (Norton et al., 2002; Davidson et al., 2008; Drillis et al., 1964; Sheets et al., 2010; Pandis & Bull, 2017). The data for each segment are available from cadaver studies that were able to determine individual segment densities based on weight and immersion tests (Table 5.4). Density has been shown to vary along a segment's length (Wei & Jensen, 1995) but the effects of implementing density profiles show alterations

in BSPs that rarely surpass 10%. The threshold error at which point this becomes important in musculoskeletal analysis is not known (Section 5.1), and therefore segment-specific density profiles were not considered in this study.

Table 5.4 – Segment density values provided in the literature from cadaveric studies. All values given as x10 $^{-3}$ kg/cm^3. The weighted averages were based on the number of cadavers/subjects for each study (segment densities) and number of studies reporting each segment (whole body density)

Segment	Dempster (1955)	Harless (1860)	Drillis & Contini (1966)	Clauser et al. (1969)	Chandler et al. (1975)	Weighted averages
Head	1.11	1.11	-	1.07	1.06	1.09
Upper arm	1.07	1.08	1.09	1.06	1	1.07
Forearm	1.13	1.10	1.13	1.10	1.05	1.11
Hand	1.17	1.11	1.15	1.11	1.08	1.13
Trunk	1.03	-	-	1.02	0.85	0.99
Thigh	1.05	1.06	1.09	1.04	1.02	1.06
Shank	1.09	1.09	1.10	1.08	1.07	1.09
Foot	1.09	1.09	1.11	1.08	1.07	1.09
					Weighted average for whole body	1.08

Given the assumption of a uniform density within a segment, it could be argued that it is valid to use similar values across the body. However, due to the unique combination of bone, muscle and fat in each segment (Winter, 2009), it cannot be assumed that the whole body has a uniform density. Cadaveric studies, such as those conducted by Harless (1860) and Dempster (1955), established density values for individual segments which were used to produce approximate profiles relating whole-body and segment density (Figure 5.4). It should be noted that these studies were based on small numbers of cadavers (seven and eight respectively) which would not have enough statistical power to determine the linear relationships that they present. This was, however, used to briefly determine if altering segment density based on whole-body density would be necessary.

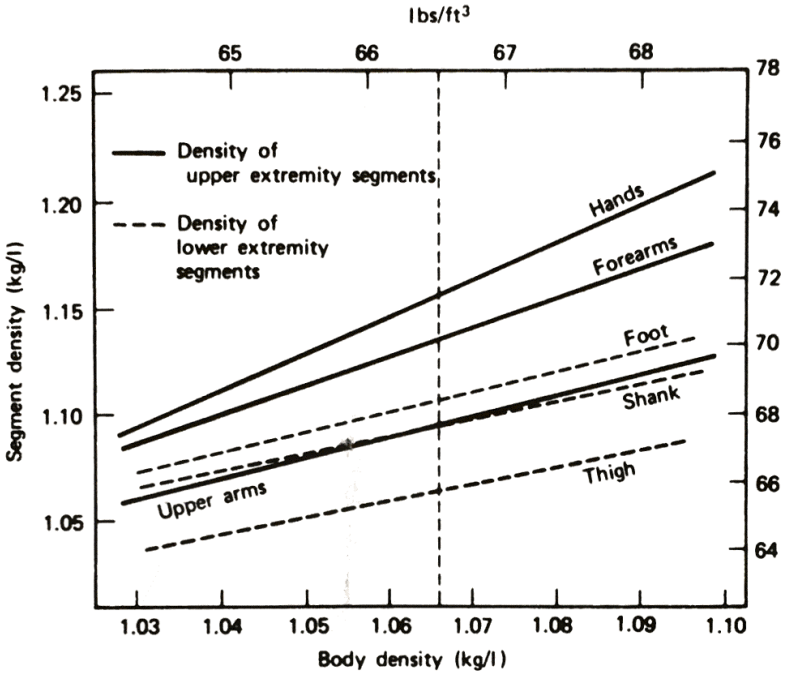


Figure 5.4 - Density of limb segments as a function of mean body density. Image reproduced with permission from Winters, 1990

Whole-body density (d) is calculated as:

$$d = 0.69 + 0.9c \ \frac{kg}{l} \quad \text{(Winter, 2009)} \qquad \text{Equation 5.1}$$

$$\text{where } c = \frac{h}{w^{\frac{1}{3}}},\ h = \text{height in metres and } w = \text{mass in kg.}$$

The work of Behnke et al. (1995) was used to quantify whole-body density values of 1.02 x10^{-3} and 1.097 x10^{-3} kg/cm^3 (minimum and maximum whole body densities of the sample population). Using these values and the profiles shown in Figure 5.4, segment densities for the thigh (used here as an example) were approximated at 1.04 x10^{-3} and 1.08 x10^{-3} kg/cm^3. Segment mass was calculated by taking an average volume for the thigh from the study by Harless (1860) of 4780.3 cm^3 and multiplying it by the approximated segment densities. This gave mass approximations of 4.97 and 5.16 kg respectively, a range of 3.7%. Considering the sensitivity studies previously reviewed, this would be of little significance when looking at the outputs of MSK

modelling. In light of this, subject-specific whole body and segment densities were not considered for this study.

Despite not considering subject- or segment-specific density profiles, it may still have been sensible to assign segment-specific densities for general use. Looking at the values in Table 5.4 and the profiles in Figure 5.4, it appears that upper limbs are denser than lower limbs. However, this may be a consequence of the cadaver studies typically using older subjects that disproportionally lose lower limb mass due to ageing prior to losing upper limb mass. Chandler et al. (1975) makes the point that trunk densities of the cadaver will always be lower due to large amounts of air, a result of organ collapse after death. This may also be the case for all cadaver studies, as the values for the trunk shown in Table 5.4 are much lower than other segments. Due to the insignificance of subject-specific density values, it was thought sensible to perform a similar analysis on whether or not to use average segment density values. Taking an overall mean of the segment densities (final row in Table 5.4) gave 1.08 $x10^{-3}$ kg/cm^3. If this was used as the density for all segments, compared to using segment-specific values, then the differences in segment mass could be calculated. Using the values in Table 5.4, the largest possible difference between segment-specific density and the overall mean of 1.08 $x10^{-3}$ kg/cm^3 would be relative to the largest value reported for the hand: 1.17 x 10^{-3} kg/cm^3. The trunk was omitted due to the potential errors discussed above. Using Harless for volumetric data (383.1 cm^3), this gave two values for hand segment mass of 0.41 and 0.45 kg; a difference of 9% in segment masses. This, unlike the previous calculations for subject-specific density profiles, may prove significant. The subsequent effects on centre of mass and inertial properties should also not be ignored, and could drastically change with this variation in density. For the sake of this study, the mean values for segment-specific density were used (values in the final column of Table 5.4).

5.3.2.2 Segment mass

To calculate segment mass, the volume of each scan must be determined. This was performed using a combination of Delaunay Triangulation and slice separation thickness. As shown in Figure 5.3, the scans are built from layered data. Each layer consists of a 2D surface profile (Figure 5.5) which was sectioned into smaller shapes using Delaunay Triangulation.

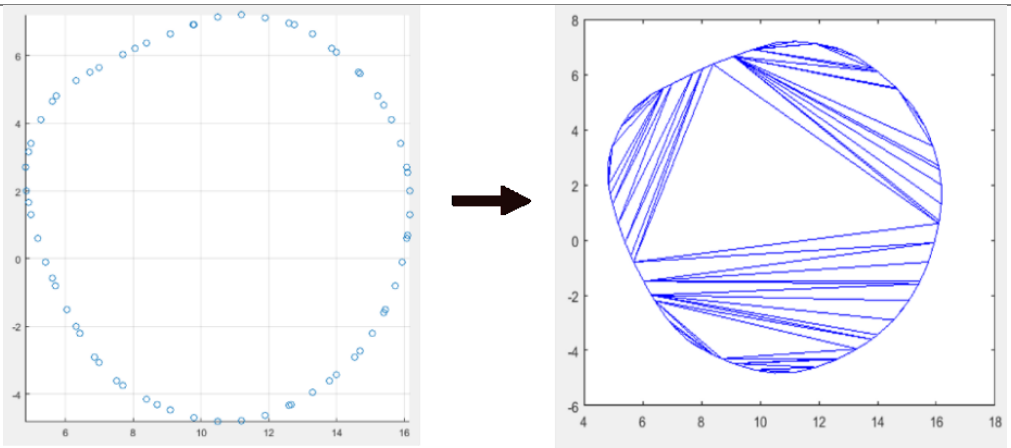


Figure 5.5- Example scan layer (left) and after triangulation (right)

The area and centroids were calculated for each triangle:

$$Area = \sqrt{s * (s-a) * (s-b) * (s-c)} \qquad \text{Equation 5.2}$$

Where: $s = \frac{1}{2} * (a + b + c)$, and a, b, and c are the lengths of the three sides of each triangle.

$$Centroid_{x,y} = (\frac{1}{3}(x_1 + x_2 + x_3), \frac{1}{3}(y_1 + y_2 + y_3) \qquad \text{Equation 5.3}$$

Where $x_{1,2,3}$ and $y_{1,2,3}$ are the x and y coordinates for the three vertices of each triangle from the global axis.

The distance between layers (Figure 5.6) was used to determine the volume of each small mass. This distance was typically between 0.6 and 0.7 cm. This was multiplied by the relevant density as given in Table 5.4 to give individual triangle mass which was summed per layer, and then per segment, to give mass.

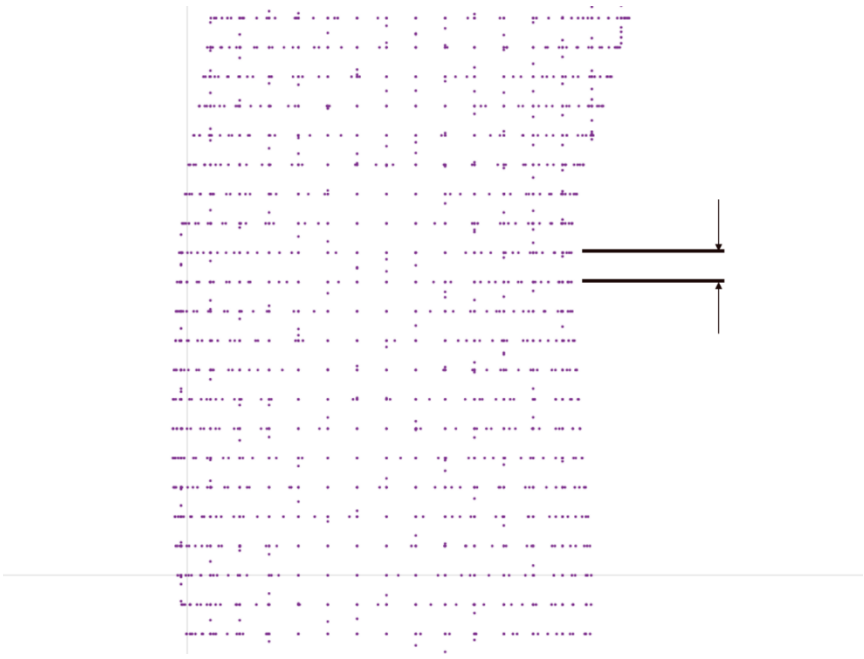


Figure 5.6 – Example scan data for the shank. Arrows indicate the separation distance used to determine layer thickness. This distance is used to represent the mass either side of each 2D profile. Separation values are consistent throughout individual scans, but varied between subjects. The first and last layers within a segment were assigned half this distance

Previous studies calculating whole-body mass found significant differences between a subject's measured mass and the sum of the individual, estimated segment masses (Sheets et al., 2010). Sheets and colleagues therefore applied a correction factor based on the differences between measured and estimated, however this may be difficult to justify on a whole-body scale. A similar adjustment was implemented by Clauser et al. (1969), who noted that the sum of a limbs' parts did not always equal the measured limb weight. This was attributed to a loss of bodily fluids during cadaver dismemberment. Clauser implemented an adjustment by increasing a segment's measured weight so that each maintained its weight ratio within the whole limb. For instance, a difference of 100 g between leg weight and the sum of foot, calf and thigh weights would be distributed by values equal to the relevant ratios for each segment. If the ratio was 40:40:20, then the thigh, calf and foot weights would be increased by 40, 40 and 20 g respectively.

The distribution of mass as a consequence of body mass estimation error amongst the segments may not be this straight forward. Without proper validation, the

amount by which each segment was over- or underestimated cannot be quantified. This is a problem for 3D scanning of living subjects as the actual weights of each limb cannot be empirically determined. A similar method to that of Clauser et al. (1969) was therefore applied for the current study, whereby the ratio of estimated segment weights to estimated whole body weight was used to distribute mass differences between measured and estimated whole body weight.

5.3.2.3 Centre of mass

Having determined the centroid of each 2D triangle in the previous section, it was then possible to find the centroid of the whole 2D layer. As a uniform density is used across each segment, centre of area was equivalent to the CoM. The centroid of the layer was calculated as:

$$x_{layer}, y_{layer} = \frac{\sum_{i=1}^{n} A_i * (x_i, y_i)}{layer\ area} \qquad \text{Equation 5.4}$$

Where A_i = individual triangle area, n = number of elements in triangulation and (x_i, y_i) = distances from global axis to triangle centroid. An iteration of Equation 5.4 can then be used to calculate the coordinates for the CoM of the whole segment relative to the global axis:

$$x_{segment}, y_{segment}, z_{segment} = \frac{\sum_{layer=1}^{n} M_{layer} * (x_{layer}, y_{layer}, z_{layer})}{layer\ mass} \qquad \text{Equation 5.5}$$

Where M_{layer} = mass of the whole layer and z_{layer} = the height of the layer relative to the global axis. Previously, studies that calculated BSPs were restricted to single plane data (Durkin & Dowling, 2003) and therefore the CoM could only be given as an x and y coordinate, or relative to a proximal/distal joint. Using 3D scanning, the CoM can be calculated as a 3D coordinate which takes into account mass distribution in the anterior-posterior direction in addition to existing information for the medial-lateral and superior-inferior directions. One restriction of existing regression tables is that the proximal/distal segment ends are aligned with the centre of mass, despite evidence to the contrary (Dumas et al., 2007; McConville et al., 1980). The method presented here therefore provides a more anatomically-correct CoM position.

5.3.2.4 Defining centre of mass

Most studies define the relative position of the CoM using the proximal joint, as was done for this study. The trunk is typically sectioned as a single segment, taking the Suprasternale as the proximal end and a plane through both hip joint centres as the distal end. The hand is defined using two methods: either considering the whole hand (including fingers) and defining the CoM accordingly or only using the mass which takes into account the carpal and metacarpals bones. De Leva (1996), through adjustments to Zatsiorksy-Selyanov's (1990) work, and Clauser et al. (1969) give CoM values in terms of the whole hand, but report it as a percentage of the carpal/metacarpal region. Similarly, Harless (1860) calculates it for the whole hand but reports it as a percentage of the whole length. As the hand is a complex structure, with a large number of articulating sections, it is difficult to define one CoM value. Harless (1860) discusses the importance of segment separation, stating that the "boundary of the mass must coincide with that of the lever arm"(Harless, 1860). For the current study, due to the limitations in resolution of the scanning method, the hand will be considered as a whole segment. Future work should look to include more detailed BSPs for the individual joints of the hand.

5.3.2.5 Inertial tensor

Inertia describes the resistance to a change in body motion, and mass moment of inertia describes resistance of a body to being rotated (Drillis & Contini, 1966). The moments of inertia about all axes are important BSPs, particularly when considering segment inertia as the main contributor to joint power requirements during movements with no additional external forces or high accelerations.

Firstly, the moment of inertia for each triangulation element about the layer CoM is found:

$$I_{element,CoM} = \sum_{i=1}^{n} m_i * ({x_i}^2 + {z_i}^2, {z_i}^2 + {y_i}^2, {x_i}^2 + {y_i}^2) \qquad \text{Equation 5.6}$$

Where i = triangle (element), m_i = triangle mass, and $({x_i}^2 + {z_i}^2, {z_i}^2 + {y_i}^2, {x_i}^2 + {y_i}^2)$ indicates the Euclidean distances between the triangle centroid and the layer y, x

and z axes, whose origins all pass through the layer centroid. As each triangle lies within the same x-y plane as the centroid, z distances are equal to 0. The parallel axis theorem determines the moment of inertia of the layer about an arbitrary axis, in this case the global system as determined by the 3D scanner calibration:

$$I_{layer,axis} = (\sum_{element=1}^{n} I_{element,CoM}) + m_{layer} * (x_{layer}{}^2 + z_{layer}{}^2, z_{layer}{}^2 + y_{layer}{}^2, x_{layer}{}^2 + y_{element}{}^2) \quad \text{Equation 5.7}$$

Where m_{layer} = total mass of the layer and $(x_{layer}{}^2 + z_{layer}{}^2, z_{layer}{}^2 + y_{layer}{}^2, x_{layer}{}^2 + y_{element}{}^2)$ indicates the Euclidean distances between the layer centroid and the y, x and z axes of the arbitrary system. An iteration of the parallel axis theorem then determines the segment's inertial tensor around its own CoM. As the sum of the moments of inertia of individual layers (which make up the segment), about an arbitrary axis, equals the whole segment's moments of inertia about an arbitrary axis, then:

$$I_{segment,CoM} = I_{segment,axis} - m_{segment} * (x_{segment}{}^2 + z_{segment}{}^2, z_{segment}{}^2 + y_{segment}{}^2, x_{segment}{}^2 + y_{segment}{}^2) \quad \text{Equation 5.8}$$

where $$I_{segment,axis} = \sum_{layer=1}^{n} I_{layer}$$

And where $m_{segment}$ = total segment mass and $(x_{segment}{}^2 + z_{segment}{}^2, z_{segment}{}^2 + y_{segment}{}^2, x_{segment}{}^2 + y_{segment}{}^2)$ indicates the Euclidean distances between the segment CoM and the y, x and z axes, respectively, of the arbitrary system.

5.3.2.6 Rotation of segments

Participants were asked to stand in the anatomical stance. This meant that the arm segments (upper arm, forearm and hand) were not positioned in line with the global vertical axis. Therefore, calculated inertial tensors must be aligned with the appropriate anatomical coordinate system. For example, this meant aligning the long axis of the segment with the vertical axis of the global system (Figure 5.7). Calculating the angle

of each segment relative to the vertical axis (Figure 5.8) allowed this to be corrected, and transformation equations were used to re-calculate inertial properties. This gave the inertial tensor aligned with the newly positioned segments.

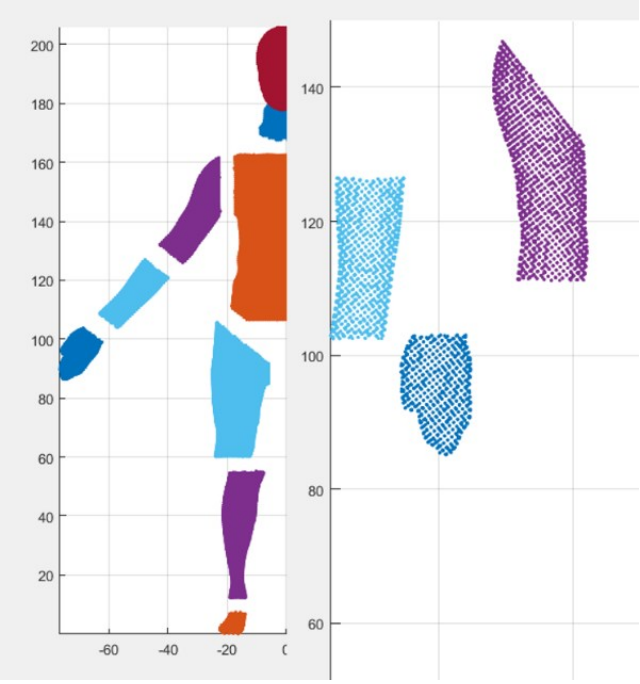
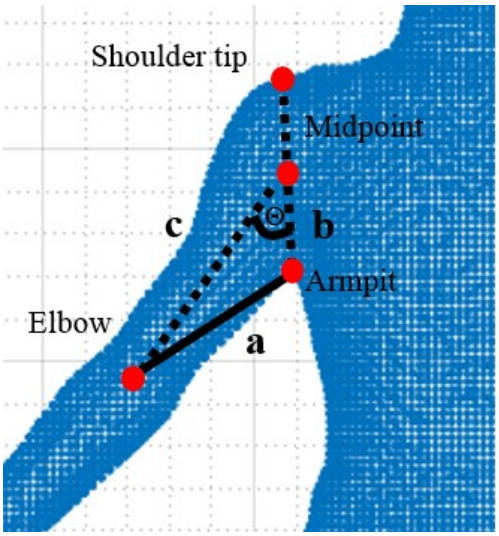


Figure 5.7 - Example scan showing original positions of limb segments (left) and rotated segments after correction (right)



1)

$$Midpoint = \frac{Shoulder\ tip + Armpit}{2}$$

2)

$$\Theta = cos^{-1}\left(\frac{b^2 + c^2 - a^2}{2 * b * c}\right)$$

Figure 5.8 - Calculation of arm angle to the body: 1 – the midpoint between the shoulder tip and armpit forms a triangle between the armpit and elbow, 2 – the cosine rule is used to find the angle between the shoulder tip-armpit plane and arm long axis

Transformations were only required in the z (superior-inferior) and x (medial-lateral) axes:

$$I_{x'x'} = \frac{I_{xx}+I_{zz}}{2} - \frac{I_{xx}-I_{zz}}{2} cos2\theta + I_{xz} sin2\theta \quad \text{Equation 5.9}$$

$$I_{z'z'} = \frac{I_{xx}+I_{zz}}{2} + \frac{I_{xx}-I_{zz}}{2} cos2\theta - I_{xz} sin2\theta \quad \text{Equation 5.10}$$

Where I_{xx} = segment original moment of inertia about CoM, $I_{x'x'}$ = corrected moment of inertia about CoM and θ = angle of segment relative to vertical axis. I_{xz} = product of inertia, and is calculated as follows:

$$I_{xz,layer,axis} = (\sum_{element=1}^{n} I_{element,CoM}) + m_{layer} * (x_{layer} * z_{layer}) \quad \text{Equation 5.11}$$

$$I_{xz=} I_{segment,CoM} = (\sum_{layer=1}^{n} I_{xz,layer,axis}) - m_{segment}(x_{segment} * z_{segment}) \quad \text{Equation 5.12}$$

5.3.3 Validation using geometric modelling

Comparisons between the results in this current study and the literature does not validate the subject-specific uses of such methods. Using literature regression values could achieve this, however, potential errors from using different populations (see Section 5.1) would make direct comparisons difficult. As discussed in Section 5.2, Hanavan (1964) developed a mathematical model of the human body for calculation of BSPs based on subject-specific anthropometric measurements and geometric modelling. This has previously been used to validate bespoke BSP calculations (Pandis, 2013; Durkin & Dowling, 2003) and will serve as an approximate validation for this study.

Three changes were made to the geometric model for use in the current study:

1. As the subjects were asked to stand in the anatomical stance with hands held open, an elliptical cylinder was used to model the hand, as validated by Pandis (2013);

2. The trunk was modelled as a singular elliptical cylinder as opposed to separating the segment into upper and lower parts;
3. Due to difficulties in obtaining an accurate circumference of the most proximal part of the arm from the 3D scan data, the upper arm was modelled as a cylinder. Pandis (2013) also experienced this issue and found the cylinder to be suitable for mass and CoM but differed for moments of inertia.

All other shapes and anthropometric measurements required are set out in Table 5.2. It is worth noting that an error was found within the Hanavan (1964) model calculations. Where moments of inertia are calculated for the trunk elliptical cylinder, a factor of 1/12 appears to be missing which gives inertial properties at approximately 10 times the expected values. This was corrected and implemented into the current analysis. As no anthropometric measurements were taken, dimensions for the geometric modelling were taken directly from the scan data.

5.3.4 Group-specific regression

The presented scanning method allows the construction of group-specific regression tables, without the need for time-consuming data collection. A sub-study investigated any changes in regression error when the base-population used to construct the regression tables had specific characteristics. Table 5.5 contains a detailed breakdown of the subjects used in the initial scanning study, with further groups based on age, gender and BMI. These groupings were based on BSP variations found to be significant in the literature (Section 5.1). Each BSP will be calculated in terms of the subject's height and/or weight, and mean values for each group will constitute that population's regression coefficients.

Table 5.5 - Population groups used to determine success of bespoke regression tables. * = Based on ranges given by the NHS body mass index (BMI) scale (NHS choices, 2016)

Group	Details	Number of subjects
All	All subjects scanned	95
Male (M)	All male subjects	62
Female (F)	All female subjects	33
Young (Y)	Subjects aged 18 to 35	70
Middle-Aged (MA)	Subjects aged 36 to 65	23
Elderly (E)	Subjects aged > 65	2
Young Males (YM)	Male subjects aged between 18 and 35	46
Young Females (YF)	Female subjects aged between 18 and 35	24
Middle-Aged Males (MAM)	Male subjects aged between 36 and 65	14
Middle-Aged Females (MAF)	Female subjects aged between 36 and 65	9
BMI – underweight (BMIU)	BMI <18.5*	1
BMI – healthy (BMIH)	BMI in the range 18.5-24.9*	64
BMI – overweight (BMIO)	BMI in the range 25-29.9*	25
BMI – obese (BMIOB)	BMI in the range 30-39.9*	5

5.3.5 Statistical tests

As a full validation of the 3D scanner volumetric data was not possible for this study, estimated segment volumes were compared with previous cadaver studies that performed immersion tests (Harless, 1860; Drillis & Contini, 1966; Clauser et al., 1969; Dempster, 1955; Chandler et al., 1975). All BSPs, barring inertial characteristics, are presented relative to subject weight or height in order to compare to literature values. Independent t-tests were used to determine statistical differences between the means of the current values and the literature, where available. Paired-sample t-tests were used to compare measured weight to estimated body weight and for comparisons against geometric modelling. Inertial characteristics are presented as absolute values. Reviewed

studies either did not present absolute inertial values (De Leva, 1996), instead presenting radii of gyration relative to height, or failed to specify segment axes (Dempster, 1955; Drillis & Contini, 1966) and therefore literature comparisons were deemed irrelevant for moments of inertia about the CoM.

For group-specific regression: normalised root-mean-square-errors (NRMSE) were used to compare BSPs calculated through regression and the bespoke values of the 3D scanning method. This was performed for all groups shown in Table 5.5 and as a general measure, for a regression table built via the data of every subject ('All'). BSPs will be grouped together (for instance, all segment masses for each population are grouped together) for each specific population and paired-sample t-tests of NRMSEs will determine the differences in estimated BSPs between regression tables.

5.4 Results

Once all 95 subjects (details given in Table 5.6) were scanned and bespoke BSPs calculated, statistical comparisons were made to literature values for segment volume, mass and CoM. To maintain fair comparisons, only studies using similar segment boundary designations to that of the current study have been included.

Table 5.6 - Subject data

Sex (n)	Age (SD, min, max), years	Height (SD, min, max), m	Body mass (SD, min, max), kg
Male (62)	34 (± 13, 18, 71)	1.78 (± 0.07, 1.62, 1.92)	78.2 (± 10.5, 59,2, 106.7)
Female (33)	32 (±10, 19, 64)	1.66 (± 0.06, 1,53, 1.80)	62.1 (± 9.6, 48.8, 89)

5.4.1 Segment Volume

Only the work by Dempster (1955) and Chandler et al. (1975) contained segment volume data for the whole body, allowing for comparisons between segment volumes relative to whole body volumes. Where appropriate, right and left sides have been

included as separate data points. Table 5.7 shows the comparisons for all eight segments. Standard deviations (SD) are calculated from the data of Dempster and Chandler only. This study found significantly higher volumes barring the feet and trunk which were lower and the shank which was found not be significantly different to the literature.

Table 5.7 - Comparisons between current and literature segment volumes. All literature values have been combined into one group. * ($p<0.05$) and ** ($p<0.0005$) indicate significant differences from the current study. Literature sample sizes: Dempster, n=7 (n = 14 with both right and left hand sides) and Chandler, n=6 (n = 12 with both right and left hand sides)

Segment	Study	N	Mean segment volume (% total body volume ± SD)
Feet	This study	190	0.6 (0.1)
	Literature	26	1.2 (0.2) **
Shank	This study	190	3.9 (0.6)
	Literature	26	3.9 (0.6)
Thigh	This study	190	10.4 (1.3)
	Literature	26	9.4 (1) **
Trunk	This study	95	52 (4.1)
	Literature	13	55.3 (2.9) *
Hands	This study	190	0.6 (0.1)
	Literature	26	0.5 (0.1) **
Forearm	This study	190	1.6 (0.3)
	Literature	26	1.4 (0.2) *
Upper arm	This study	190	3 (0.4)
	Literature	26	2.6 (0.3) **
Head	This study	95	7.7 (1.3)
	Literature	13	6.6 (1.2) *

5.4.2 Segment mass

The full body weight of each subject was measured prior to the scan for comparisons to estimated body weight (i.e. the sum of all segment masses). The mean estimated weight was 103.1 (± 5.7) % of the measured weight ($p<0.0005$). Table 5.8

shows the differences between the results from this study and the literature. Raw data comparisons were possible against the works of Dempster (number of cadavers, n =16), Chandler (n=12), Harless (n=3), Clarys & Marfell-Jones (n=12) and Braune & Fischer (n=6), details of which can be found in Table 5.1. Other studies (Drillis & Contini, 1966; De Leva, 1996; Jensen & Fletcher, 1994; Durkin & Dowling, 2003; Clauser et al., 1969) reported mean and standard deviation (though not all studies have SD) values only, and were included as individual data points. Standard deviation values for the literature were based on all data from the studies reviewed.

Table 5.8 - Comparisons between this study and literature segment masses. All literature data have been combined into one group and averaged. * ($p<0.05$) and ** ($p<0.0005$) indicate significant differences from the current study

Segment	Study	N	Mean segment mass (% body weight ± SD)	Mean difference (% This study)
Feet	This study	190	0.64 (0.14)	-129.65 **
	Literature	53	1.48 (0.29) **	
Shank	This study	190	4.19 (0.62)	-0.90
	Literature	54	4.23 (0.57)	
Thigh	This study	190	10.76 (1.34)	-3.44
	Literature	54	11.13 (2.62)	
Trunk	This study	95	49.59 (4.16)	-2.07
	Literature	28	50.62 (5.10)	
Hands	This study	190	0.71 (0.16)	7.87 *
	Literature	53	0.65 (0.13) *	
Forearm	This study	190	1.68 (0.33)	4.31
	Literature	54	1.61 (0.27)	
Upper arm	This study	190	3.08 (0.42)	8 **
	Literature	53	2.84 (0.34) **	
Head	This study	95	8.26 (1.36)	14.28 **
	Literature	27	7.08 (0.84) **	

Significant differences were found for the feet, hands, upper arms and head. The upper arm, hands and head were overestimations, whereas the feet values were

underestimated compared to the literature. For a visual comparison against individual studies see Figures 5.9 to 5.12; note that not all studies contained data for all segments.

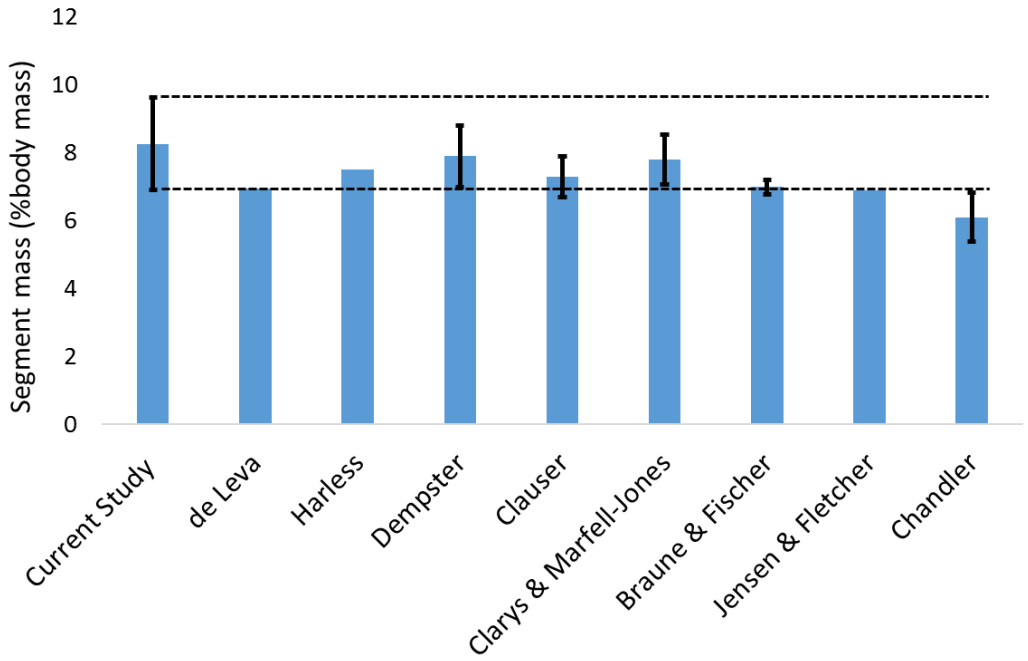


Figure 5.9 - Head mass as a percentage of whole body mass. Standard deviations added where full datasets were available in the literature. Dashed lines indicate upper and lower limits of the current study's standard deviations.

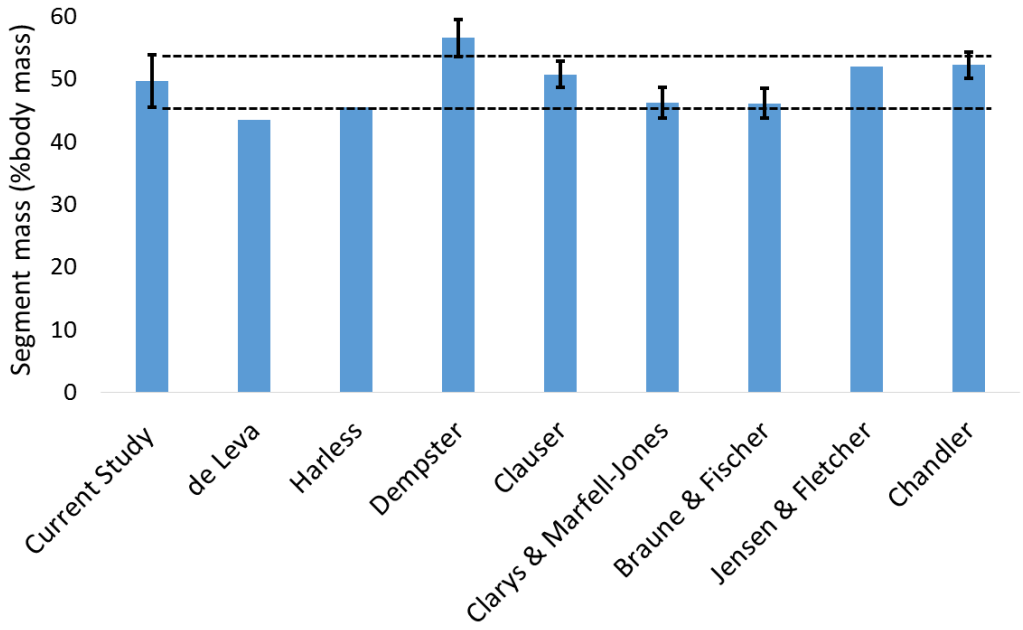


Figure 5.10 – Trunk mass as a percentage of whole body mass. Standard deviations added where full datasets were available in the literature. Dashed lines indicate upper and lower limits of the current study's standard deviations.

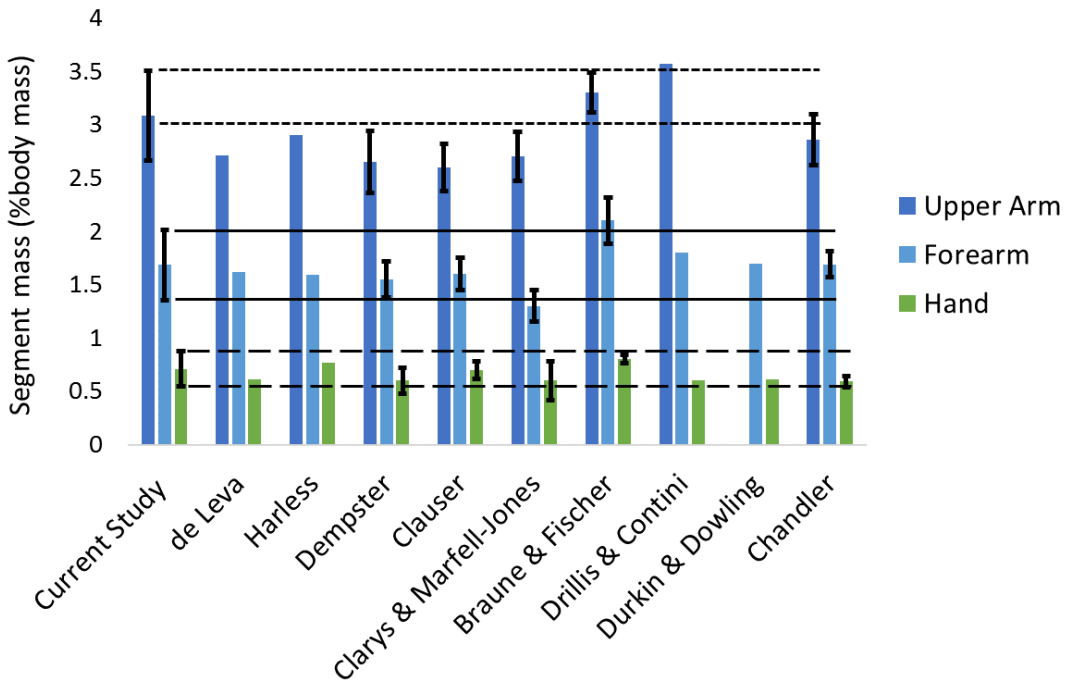


Figure 5.11 – Upper limb segment masses as percentages of whole body mass. Standard deviations added where full datasets were available in the literature. Dashed lines indicate upper and lower limits of the current study's standard deviations.

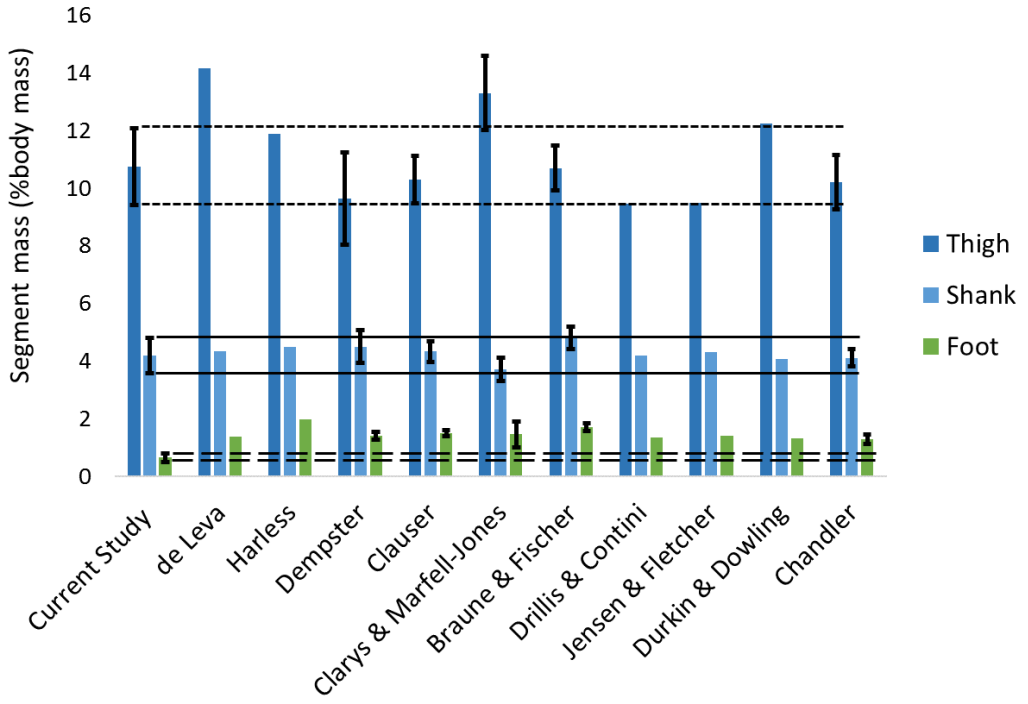


Figure 5.12 – Lower limb segment masses as percentages of whole body mass. Standard deviations added where full datasets were available in the literature. Dashed lines indicate upper and lower limits of the current study's standard deviations.

5.4.3 Centre of mass

Table 5.9 shows the differences between the current study and the literature CoM values from Chandler (1975: n=12) and Braune & Fischer (1889, n=6). Work by Dempster (1955), Harless (1860), Drillis & Contini (1964), Clauser (1969), De Leva (1996) and Durkin & Dowling (2003) reported mean and standard deviation values only, and were included as individual data points. Standard deviation for the literature was based on all data from the studies reviewed. In keeping with the segment boundary definition, CoM of the hands was presented as a percentage of the whole hand and adjusted accordingly if the literature studies stated otherwise. All values are given relative to the proximal joint except the head which is given relative to the most superior point. Figures 5.13 to 5.16 show the comparisons between 3D scanning results and the individual studies reviewed from the literature.

Table 5.9 - Comparisons between current and literature segment CoM as percentages of segment length. All literature data have been combined into one group. * ($p<0.05$) and ** ($p<0.0005$) indicate significant differences from the current study

Segment	Study	N	Mean CoM (% segment length from proximal end ± SD)	Mean difference (% Current)
Feet	This study	190	44.2 (1.8)	1
	Literature	23	43.8 (1.3)	
Shank	This study	190	37.2 (1.6)	-12.4 **
	Literature	24	41.8 (1.8) **	
Thigh	This study	190	43.1 (2.1)	4.9 **
	Literature	24	41.0 (3.3) **	
Trunk	This study	95	51.3 (1.6)	1.2
	Literature	10	50.7 (5.9)	
Hands	This study	190	41.1 (1.7)	-17.4 **
	Literature	17	48.3 (7.1) **	
Forearm	This study	190	42.8 (1.7)	2.6 *
	Literature	21	41.6 (2.1) *	
Upper arm	This study	190	39.8 (2.3)	-23.7 **
	Literature	22	49.2 (4) **	
Head	This study	95	60 (5.4)	2.7
	Literature	10	58.4 (12)	

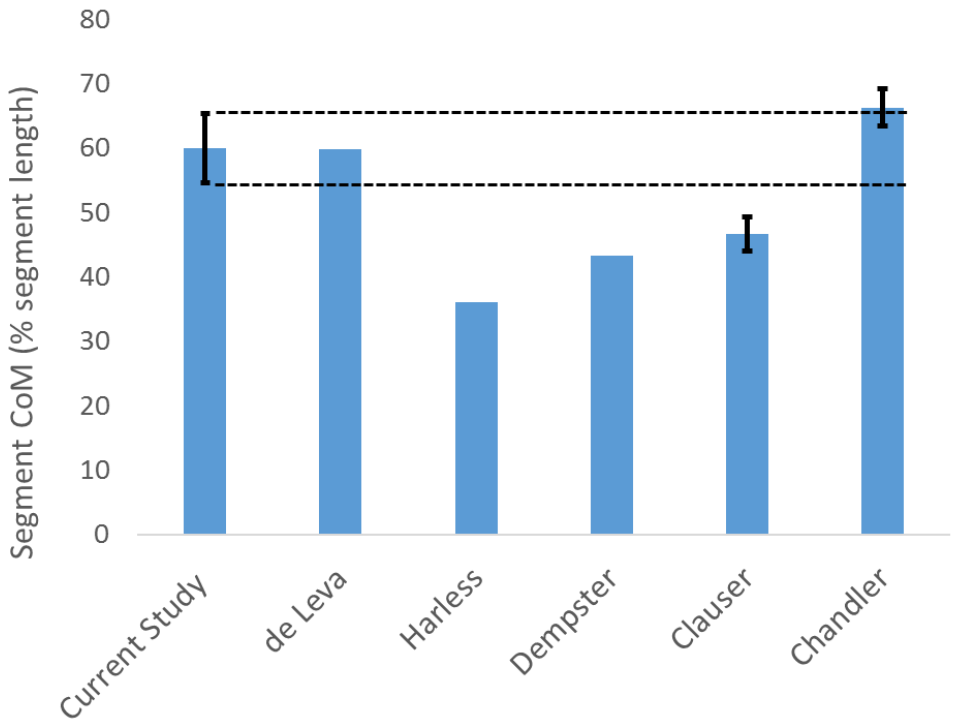


Figure 5.13 – Head centre of mass as a percentage of segment length, relative to the vertex of the skull Standard deviations added where full datasets were available in the literature. Dashed lines indicate upper and lower limits of the current study's standard deviations.

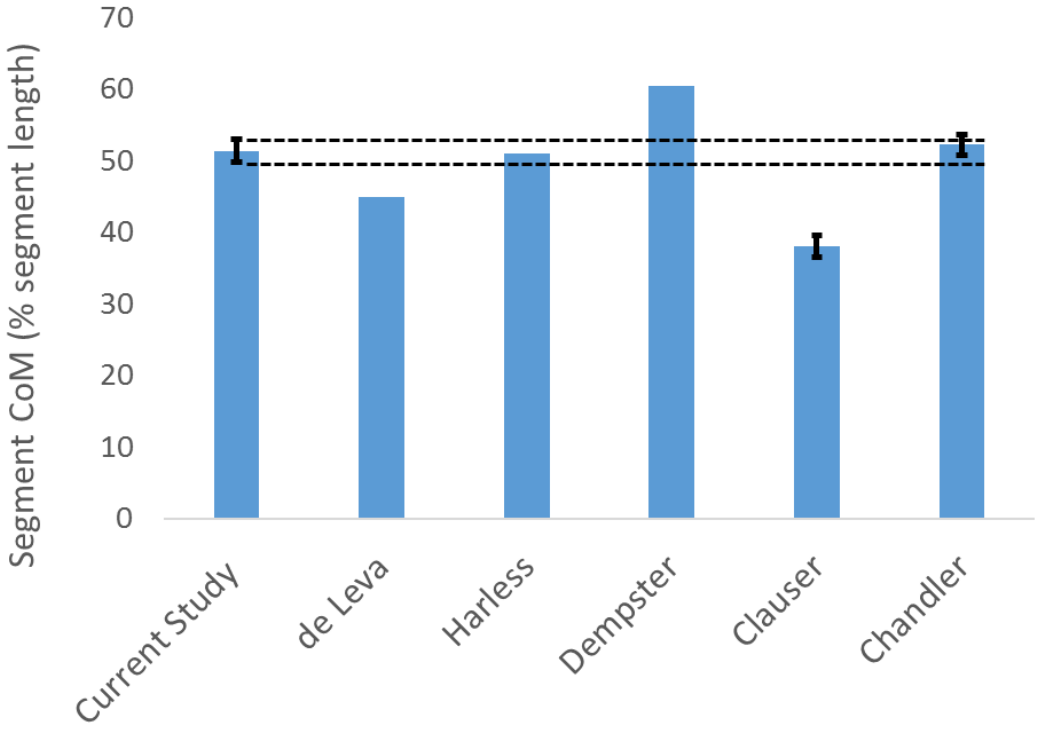


Figure 5.14 – Trunk centre of mass as a percentage of segment length, relative to the jugular notch. Standard deviations added where full datasets were available in the literature. Dashed lines indicate upper and lower limits of the current study's standard deviations

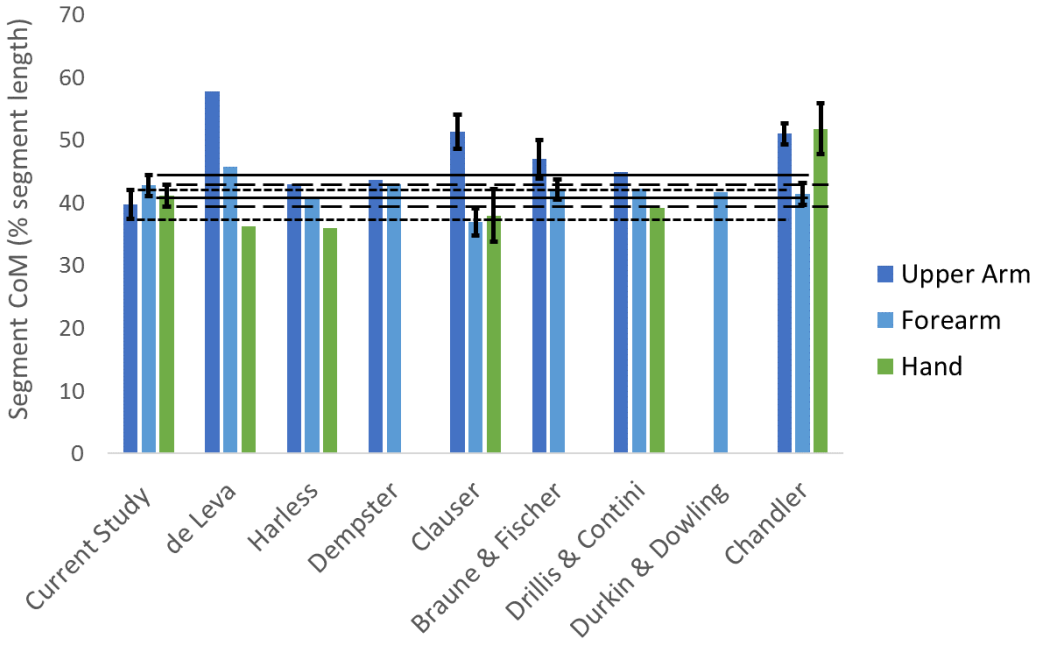


Figure 5.15 – Upper limb centres of mass as a percentage of segment length, relative to the proximal joint. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

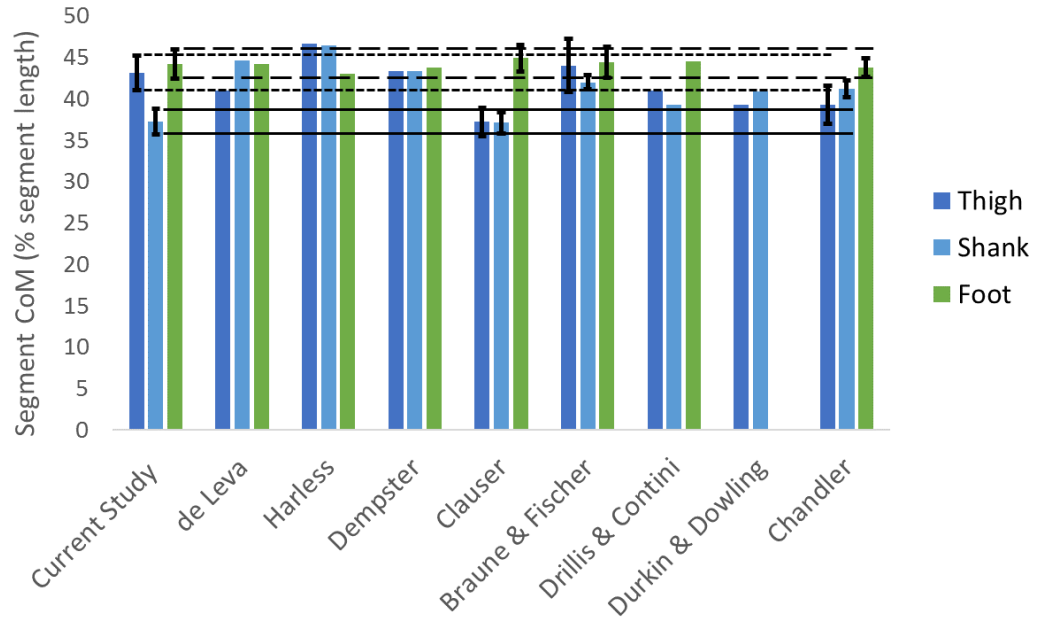


Figure 5.16 – Lower limb centres of mass as a percentage of segment length, relative to the proximal joint. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

5.4.4 Moments of inertia

All inertial properties (moments of inertia about the transverse axis = I_{xx}, the sagittal axis = I_{yy} and the coronal axis = I_{zz}) are presented as absolute values and so comparisons to studies which only present radii of gyration relative to segment length are not possible. For visual comparisons, Figures 5.17 to 5.24 show moments of inertia about the x, y and z axis for each segment's local coordinate frame against literature absolute values. All values are given about the segment CoM. All literature-derived coordinate systems matched (or were adjusted to match) the one used in this study. Additional studies by McConville et al. (1980) and Durkin & Dowling (2003), who both computed moments of inertia via medical imaging methods (stereophotometry and DEXA respectively) on living subjects, have been included as further comparisons where available.

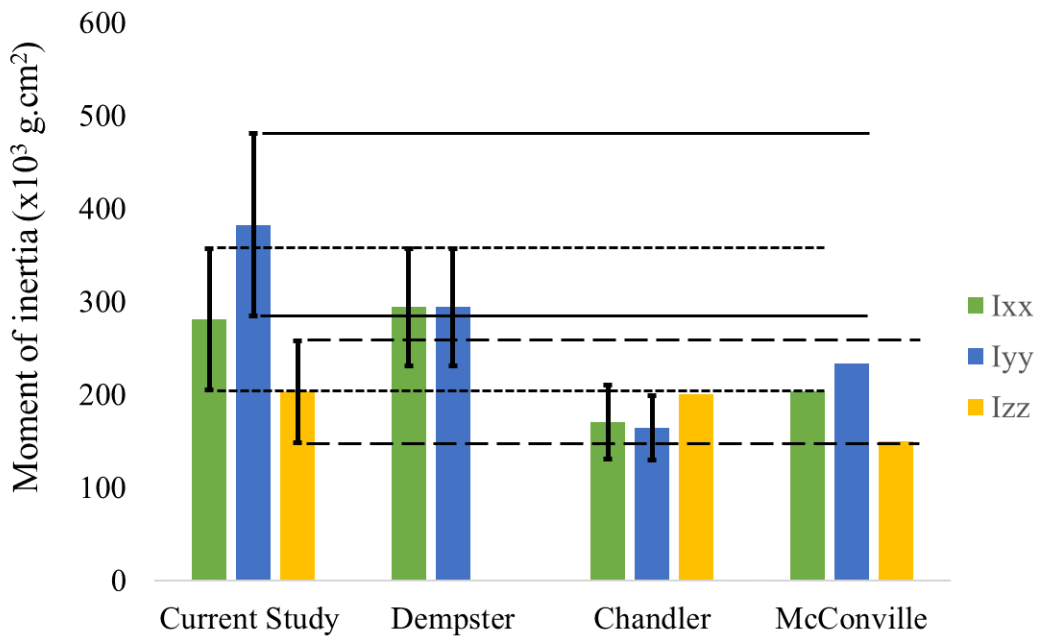


Figure 5.17 – Head inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

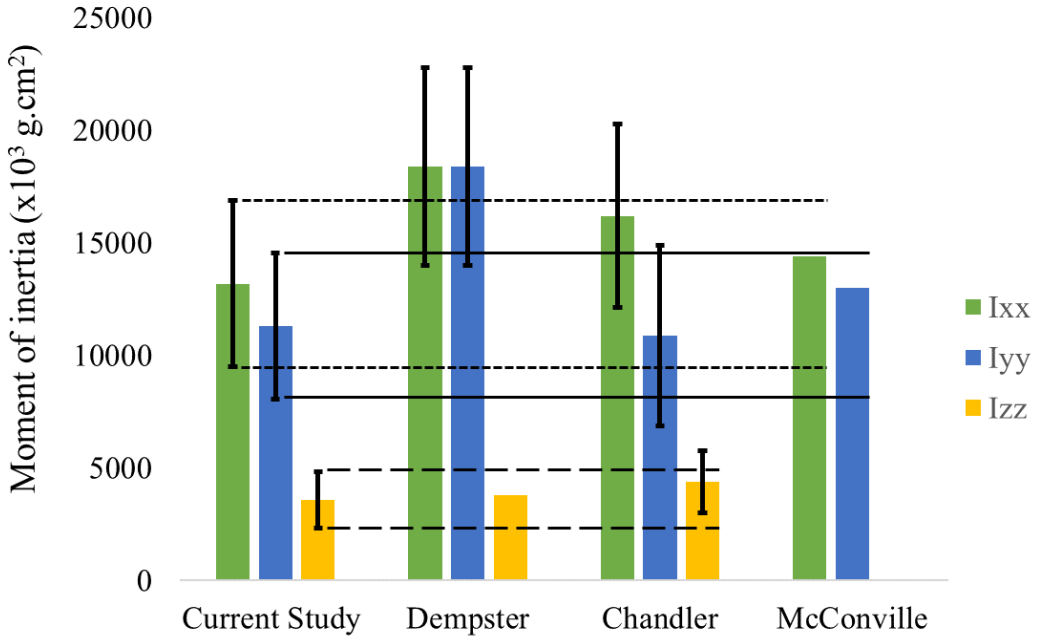


Figure 5.18 - Trunk inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

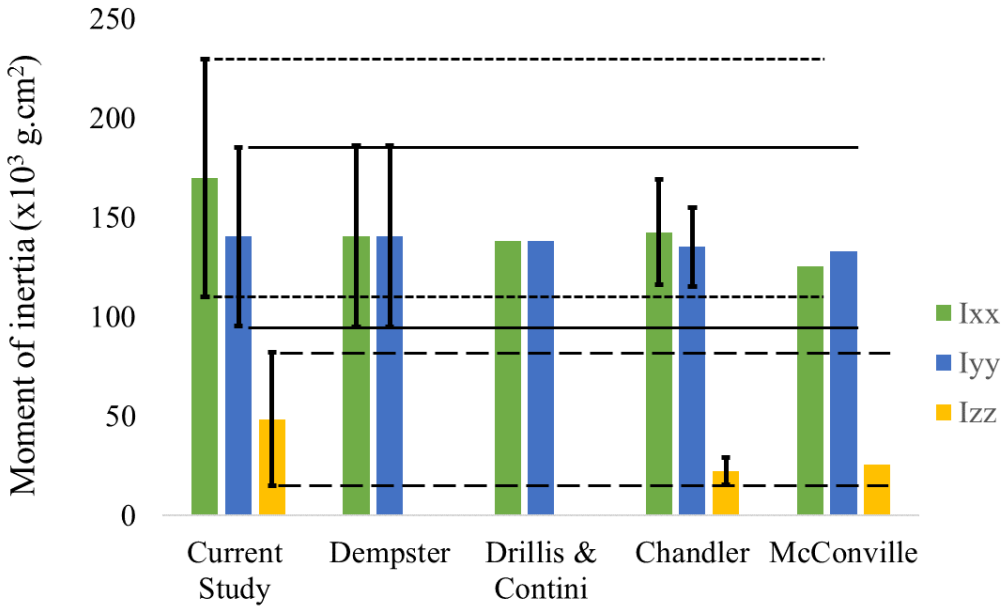


Figure 5.19 – Upper arm inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

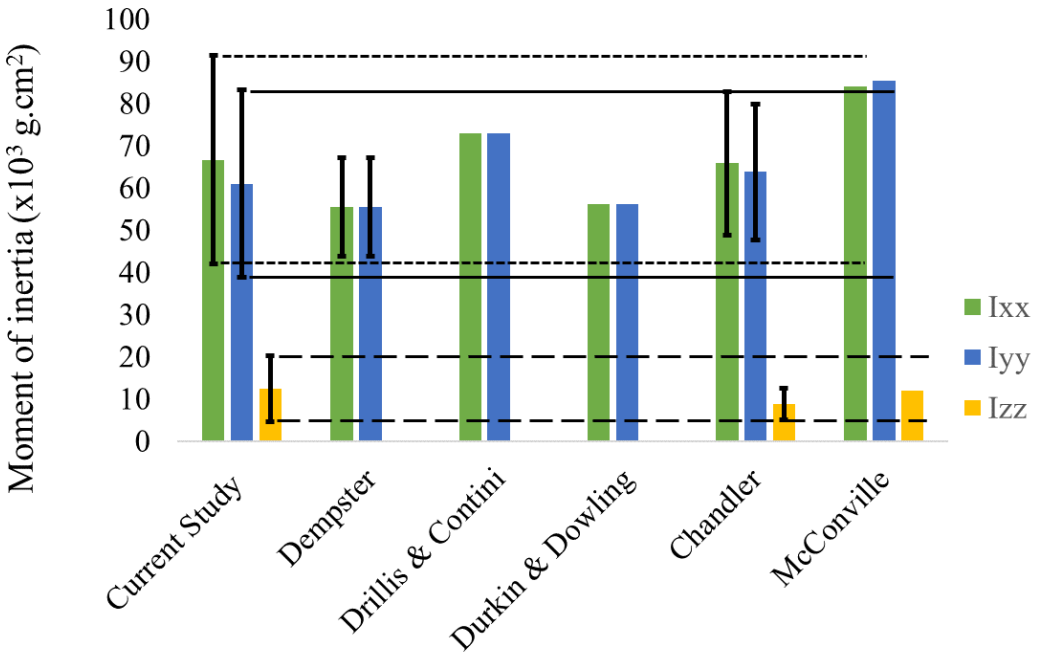


Figure 5.20 - Forearm inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

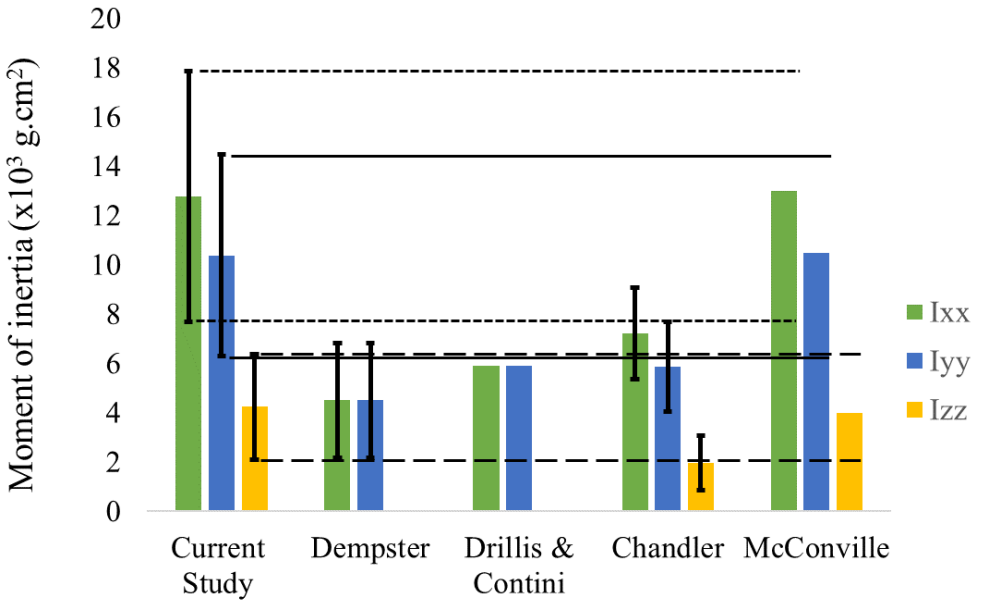


Figure 5.21 - Hand inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

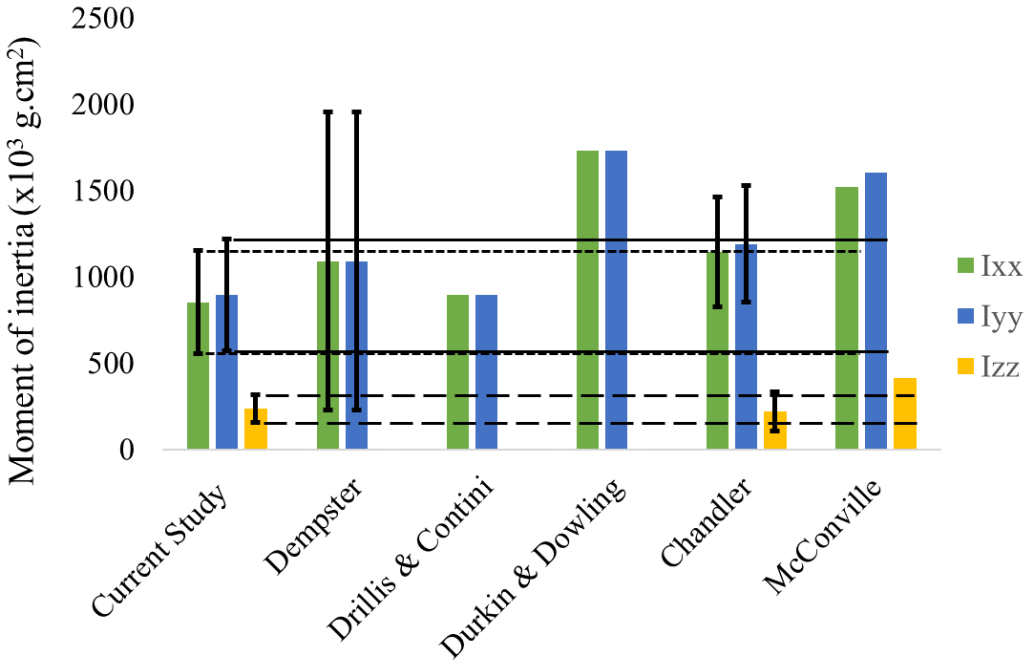


Figure 5.22 - Thigh inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

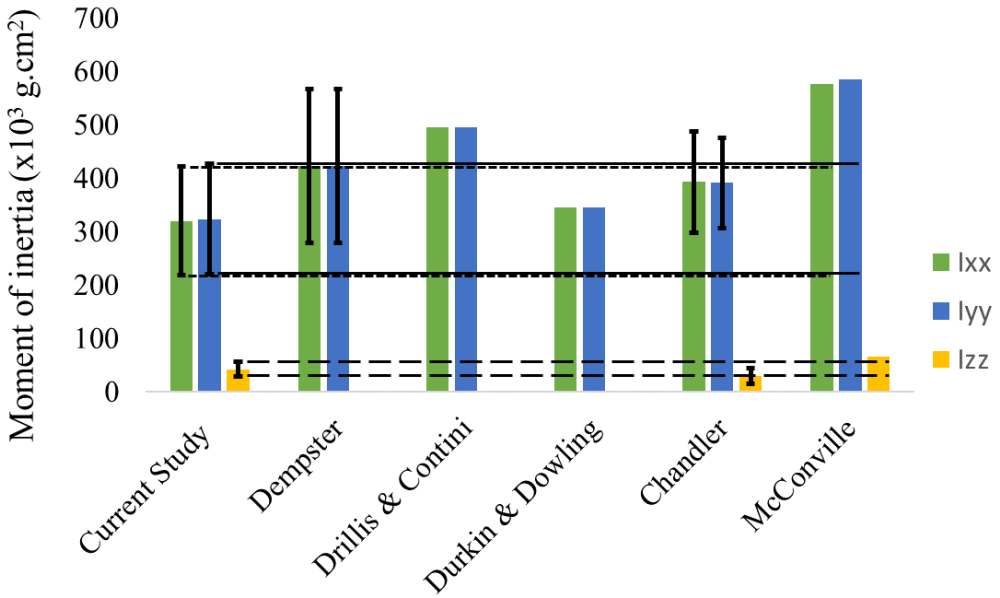


Figure 5.23 - Shank inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

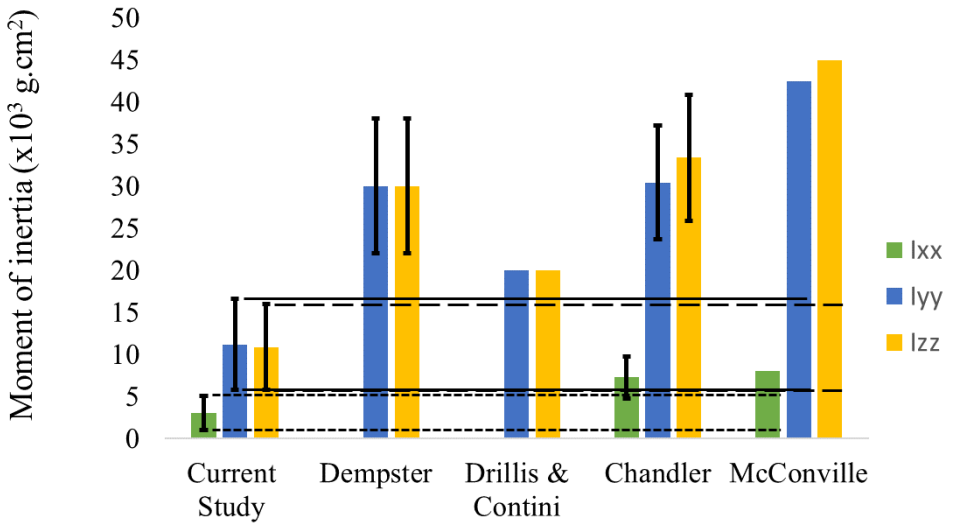


Figure 5.24 - Foot inertial properties about the centre of mass. Standard deviations added where full datasets were available in the literature. Dashed/solid lines indicate upper and lower limits of the current study's standard deviations

5.4.5 Geometric modelling

Table 5.10 (absolute) and Table 5.11 (relative to 3D scanning values) show the mean differences between the BSPs calculated through the 3D scanning method and those calculated via geometrical modelling.

Table 5.10 - Mean (± SD) absolute differences, calculated as = scanning – geometric modelling. * (p<0.05) and ** (p<0.0005) indicate significant differences between the two methods

Segment	Mass (kg)	CoM (% segment length)	I_{xx} (x 10^3g.cm^2)	I_{yy} (x 10^3g.cm^2)	I_{zz} (x 10^3g.cm^2)
Head	0.63 (0.53)**	10 (5.4)**	24.1 (37.5)**	125.6 (45.2)**	-6.8 (22)*
Trunk	0.21 (2)	1.3 (1.6)**	503 (697)**	-279.9 (752)**	-307 (257)**
Upper arm	0.24 (0.31)**	-10.2 (2.3)**	46.4 (26.6)**	16.9 (25.5)**	23.1 (27.1)**
Forearm	-0.37 (0.22)**	-2.2 (1.8)**	-24.3 (14.2)**	-30 (17.3)**	-3.5 (6.1)**
Hand	-0.35 (0.2)**	-8.9 (1.7)**	-13.1 (7.05)**	-15.9 (8.8)**	-2.6 (3) **
Thigh	1.9 (1.36)**	-3.4 (1.4)**	92 (105.5)**	133.9 (115.6)**	77.7 (62.9)**
Shank	-0.01 (0.29)	-2.5(1.7)**	-62.1 (42.3)**	-59.2 (39.8)**	2.4 (7)**
Foot	-0.01(0.08) *	13.2(2.1)**	0.4 (1.7)**	-1.1 (3.1)**	-1.4 (2.7)**

Table 5.11 - Mean (% of scanning estimate) differences between scanning and geometric modelling methods. These values represent those in Table 5.10, but relative to the scanning average values

Segment	Mass	CoM	I_{xx}	I_{yy}	I_{zz}
Head	7.63	16.66	8.57	32.83	-3.35
Trunk	0.42	2.53	3.82	-2.48	-8.54
Upper arm	7.78	-25.65	27.46	12.04	-47.72
Forearm	-21.97	-5.14	-36.27	-49.18	-28.23
Hand	-49.25	-21.63	-102.58	-153.21	-61.42
Thigh	17.66	-7.89	10.78	14.96	33.12
Shank	-0.24	-6.72	-19.41	-18.33	5.7
Foot	-1.56	29.86	13.3	-9.16	-12.84

5.4.6 Group-specific regression tables

As there were very small numbers for the Elderly (n=2), BMI-underweight (n=1), and BMI-obese (n=5) groups, these were excluded from the analysis, leaving 87 subjects in the dataset in total.

Segment mass, length, CoM and inertial properties were analysed separately for all remaining groups. All group NRMSE values for each segment and BSP are shown in Table 5.12. Segments were grouped together and paired sample t-tests were performed on the mean differences between NRMSE values and the All group.

Segment mass NRMSE values were found to be significantly smaller for groups M and YM (mean differences of 0.029 and 0.045 respectively, $p<0.05$). Segment length NRMSE values were found to be significantly smaller for the YM group only (mean difference of 0.013, $p<0.05$). The NRMSE values for all groups were less than 0.0001 for segment CoM, so paired t-tests were not deemed necessary. Inertial properties (I_{xx}, I_{yy} and I_{zz}) were grouped together, and mean differences in NRMSE for the M (0.029, $p<0.0005$), Y (0.016, $p<0.0005$), MA (0.033, $p<0.05$), YM (0.045, $p<0.05$), BMIH (0.017, $p<0.05$) and BMIO (.026, $p<0.05$) groups were found to be significantly smaller (barring MA, whose values were significantly greater) than the All group.

Table 5.12 - Normalised RMSE values for each group (abbreviations defined in Table 5.5), body segment and BSP. As centre of mass NRMSE values were all less than 0.0001, they have been omitted

Segment	BSP	All	M	F	Y	MA	YM	YF	MAM	MAF	BMIH	BMIO
Feet	Mass	0.20	0.15	0.26	0.19	0.23	0.15	0.23	0.15	0.34	0.21	0.15
	length	0.08	0.05	0.09	0.07	0.09	0.05	0.08	0.07	0.10	0.08	0.06
	I_{xx}	0.58	0.53	0.62	0.56	0.66	0.52	0.62	0.58	0.59	0.57	0.54
	I_{yy}	0.36	0.30	0.48	0.32	0.49	0.27	0.41	0.42	0.66	0.34	0.30
	I_{zz}	0.33	0.25	0.48	0.28	0.46	0.21	0.40	0.37	0.66	0.33	0.21
Shank	mass	0.15	0.13	0.16	0.13	0.17	0.13	0.12	0.13	0.25	0.13	0.17
	length	0.06	0.05	0.06	0.05	0.05	0.05	0.06	0.05	0.06	0.05	0.06
	I_{xx}	0.24	0.21	0.29	0.23	0.28	0.22	0.23	0.20	0.39	0.24	0.25
	I_{yy}	0.25	0.22	0.30	0.23	0.29	0.22	0.24	0.21	0.39	0.24	0.25
	I_{zz}	0.27	0.21	0.34	0.26	0.32	0.20	0.27	0.21	0.48	0.24	0.31
Thigh	mass	0.12	0.10	0.13	0.12	0.14	0.09	0.12	0.11	0.16	0.12	0.12
	length	0.09	0.05	0.06	0.09	0.08	0.05	0.06	0.05	0.06	0.09	0.09
	I_{xx}	0.19	0.15	0.21	0.19	0.18	0.15	0.20	0.14	0.24	0.19	0.19
	I_{yy}	0.20	0.15	0.22	0.20	0.19	0.16	0.21	0.14	0.25	0.19	0.20
	I_{zz}	0.22	0.18	0.30	0.22	0.20	0.19	0.29	0.17	0.26	0.18	0.19
Trunk	mass	0.08	0.07	0.09	0.08	0.09	0.07	0.08	0.08	0.11	0.08	0.08
	length	0.08	0.06	0.06	0.08	0.08	0.06	0.06	0.07	0.08	0.08	0.08
	I_{xx}	0.25	0.20	0.19	0.25	0.25	0.20	0.16	0.20	0.23	0.26	0.24
	I_{yy}	0.26	0.21	0.19	0.26	0.26	0.21	0.17	0.22	0.23	0.27	0.25
	I_{zz}	0.25	0.24	0.22	0.23	0.28	0.18	0.21	0.29	0.26	0.19	0.20
Upper arm	mass	0.14	0.13	0.13	0.13	0.16	0.12	0.12	0.17	0.14	0.13	0.13
	length	0.09	0.08	0.09	0.08	0.10	0.08	0.09	0.08	0.11	0.08	0.09
	I_{xx}	0.26	0.26	0.26	0.25	0.30	0.25	0.24	0.29	0.30	0.25	0.27
	I_{yy}	0.27	0.26	0.27	0.27	0.28	0.26	0.26	0.26	0.27	0.24	0.29
	I_{zz}	0.59	0.56	0.50	0.53	0.75	0.50	0.46	0.75	0.59	0.53	0.53
Forearm	mass	0.20	0.18	0.20	0.18	0.22	0.16	0.20	0.21	0.21	0.18	0.18
	length	0.08	0.08	0.09	0.08	0.07	0.08	0.09	0.07	0.07	0.07	0.09
	I_{xx}	0.30	0.27	0.35	0.28	0.32	0.26	0.35	0.30	0.35	0.28	0.28
	I_{yy}	0.30	0.28	0.35	0.29	0.32	0.26	0.35	0.29	0.36	0.29	0.27
	I_{zz}	0.54	0.52	0.47	0.54	0.57	0.52	0.49	0.52	0.42	0.46	0.52
Hands	mass	0.23	0.23	0.21	0.22	0.24	0.21	0.19	0.23	0.24	0.23	0.20
	length	0.08	0.08	0.07	0.07	0.09	0.07	0.07	0.09	0.09	0.07	0.08
	I_{xx}	0.35	0.35	0.30	0.31	0.37	0.31	0.29	0.37	0.32	0.33	0.33
	I_{yy}	0.35	0.35	0.30	0.31	0.37	0.30	0.30	0.39	0.31	0.32	0.32
	I_{zz}	0.44	0.44	0.38	0.37	0.52	0.36	0.36	0.52	0.43	0.40	0.46
Head	mass	0.17	0.16	0.18	0.18	0.12	0.16	0.21	0.12	0.05	0.16	0.12
	length	0.08	0.09	0.07	0.09	0.08	0.09	0.07	0.09	0.04	0.09	0.06
	I_{xx}	0.26	0.26	0.24	0.27	0.22	0.27	0.28	0.21	0.13	0.28	0.20
	I_{yy}	0.27	0.26	0.26	0.28	0.25	0.26	0.30	0.21	0.14	0.29	0.19
	I_{zz}	0.31	0.28	0.32	0.31	0.30	0.28	0.36	0.23	0.18	0.33	0.23

5.5 Discussion

Only one previous study has been found which used the Kinect to estimate BSPs. Kudzia (2015) scanned 21 subjects (10 male, 11 female) in a static pose whilst a handheld sensor was moved around the body. Each scan took around 30 seconds, but additional time was required for anatomical land-marking (30 minutes) and extensive post-processing (45 minutes). This compares to around 60 seconds for a scan using the method outlined in this study, followed by 10-15 minutes for manually finding landmarks on the point cloud, and less than a minute post-processing for the BSP calculation. Kudzia compared their own subject-specific BSPs with geometric modelling and ratios were compared with one other regression table from the literature. This regression table was based on living subjects, not cadavers, so no direct comparison was made to empirical data. Despite this, the results appear to closely match those of the comparisons made and provides further confirmation that the Kinect is a suitable method by which to obtain bespoke BSPs.

5.5.1 Volume

The results from the 3D scanning method outlined in this study are in good agreement with cadaveric data. The differences shown, although statistically significant, are all below 3% and can be attributed to fluid loss from the cadavers prior to immersion testing (Reid & Jensen, 1990) and the difference in demographics used for this study. Head and thigh volumes were larger than those observed in the literature data, which may be due to the methods used in determining segment boundaries in this study: hair was not accounted for, and as a result the scanner would have interpreted this data as extra head volume. While defining the thigh using estimations of the hip joint centre is a more suitable method of defining the functional segment, it may have included some distal parts of the trunk. Consequently, a reduction in trunk volume would have occurred. Feet volumes also appeared lower compared to the literature. Visual inspection of the scans showed a loss of the inferior parts of the segment due to scanner field-of-view limitations, making it difficult to produce a comprehensive view of the foot. A method presented by Kudzia (2015), whereby subjects stood on a slanted transparent block, may overcome this in future design iterations.

The number of subjects included in each study should be considered when discussing the differences between the results of this work and literature values. The study presented in this thesis has approximately seven times more data than the cadaver studies, which may impact on the statistical power. The populace of each study may also have a bearing on differences. The cadaver studies consisted mostly of elderly male subjects, whereas this study has both male and female subjects, and a large range of ages, heights and weights. Considering the loss of fluid in cadavers, and the pre-existing decreases in muscle mass with age (Chapter 2), it appears logical that the volume data presented from the 3D scanning method of living subjects would be significantly larger than that of the literature data and highlights the improvement of using 3D scanning to determine bespoke volumes.

5.5.2 Segment mass

The differences between measured whole-body weight and summation of estimated segment weights are comparable to other studies using similar scanning techniques (Sheets et al., 2010; Kudzia, 2015). A correction factor was applied to account for these differences (Section 5.3.2.2). Despite this, significant differences in segment mass between 3D scanning and the literature mass values appear to follow the same pattern as predictions for segment volume, with the feet showing the largest differences between studies. Overestimations for the head can again be attributed to the inclusion of hair in the scans. All other differences were found to be below 10%.

Looking at the upper and lower limbs, there is greater variation in mass proportions in more proximal segments compared with distal segments. This may be due to the ambiguity associated with literature-based methods of segmenting the upper arm and thigh at the shoulder/hip respectively. Despite consistencies between studies in dismembering the cadavers, there may still have been differences in the precision of segmentation due to the complexity of these joints.

5.5.3 Centre of Mass

Differences between 10 and 25% were found between the 3D scanning and literature data for CoM of shank, hands and upper arm. The differences in shank are

small considering the absolute differences were around 4%, and underestimation of the hand CoM comes as a direct result of the overestimation of mass, placing the CoM closer to the wrist. Despite the differences in foot mass, the CoM aligns well confirming that the loss of inferior segment data is the most likely cause for disparity in segment mass. This study produced significantly lower values for upper arm CoM, this can be attributed to the extra shoulder data included in the scanned data, which would move the CoM closer to the proximal end. Many cadaver studies appear to have discarded the mass around the glenohumeral head and lateral parts of the scapula, creating a purely humeral segment. However, the shoulder complex as a functional segment consists of much of the tissue surrounding the lateral parts of the scapula and as such, has been included in this study's designation of the upper arm.

5.5.4 Moments of Inertia

Few studies report experimentally-obtained moments of inertia. Unlike mass and CoM, it cannot be normalised to height and weight and relative comparisons are difficult. De Leva (1996), again through adjustments to Zatsiorksy-Selyanov's work, instead reports radii of gyration. This is defined as the distance from rotational axis to an arbitrary point where the concentrated total mass has the same moment of inertia as the original state of distributed mass (Drillis & Contini, 1966) and can be normalised to segment length. Squaring this and multiplying it by the segment mass gives the moments of inertia. However, this method will make the final values sensitive to any discrepancies in segment mass and length, and could result in large differences between estimated and regression-predicted values. Dempster (1955), Chandler et al. (1975) and Drillis & Contini (1966) all reported absolute moments of inertia values about the CoM, and these values were compared to the data in this thesis. Dempster and Drillis & Contini report values for an undefined axis, which were calculated using the parallel axis theorem based on moments of inertia taken about proximal joints. No details are given to indicate about which axis this is, however illustrations suggest it was about either the sagittal or coronal planes. This is comparable to the values reported in this thesis for I_{xx} or I_{yy} for most segments. Chandler et al. (1975) reports a set of very clear body segment images with associated local axes, each with mean values for I_{xx}, I_{yy}, and I_{zz}; the directions of which match those used in this thesis' 3D scanning study (apart from the foot which are adjusted for comparison).

There is good visual agreement between the results of this study and previous literature on the trunk and proximal limbs. Once again, inclusion of the hair on the head segments causes difficulties in obtaining comparable inertial properties. Differences between the current study and Dempster (1955) for the trunk may relate to inconsistencies in the CoM values, as Dempsters' are higher than those found by most other studies. The hands and feet do show differences, which are most likely a result of errors from mass estimation carried across into the inertial calculations.

5.5.5 Geometric modelling

In terms of absolute values, the mass data shows very little difference between the two methods. There are evidently significant differences between the sets of values, however, relative to the mean masses of each segment these differences are well within the acceptable ranges to avoid significant differences in MSK modelling outputs. The exceptions are for the forearm and hand. As discussed previously, the hand has multiple models for representation, an elliptical cylinder being used in this instance. Comparing the scanning and literature values in previous sections, the hand showed accurate mass values. Therefore, it is evident that this shape is inappropriate for the hand in this case. Geometric modelling may require a clenched fist for accurate representation using the original sphere shape.

The consistency of both groups to be significantly different can be attributed to the inability of geometric modelling to accurately represent the complex shapes of the body, something noted by others using this method (Durkin & Dowling, 2003). Table 5.11 shows these differences relative to the original scanning estimates. Many of these values are less than 20%, however there are some extreme values, particularly for the moments of inertia. Given that the scanning estimates agreed well with cadaveric literature, this table highlights the inaccuracies in using geometric modelling to represent body segments. Segment mass shows little difference, however the hand showed that geometric modelling gave values up to 50% larger than the 3D scanning.

The standard deviations in Table 5.10 should not be ignored here, as many show differences that could range to near zero. For CoM, the differences mainly highlighted segments poorly represented by geometric modelling: the head by an ellipsoid which

doesn't account for separation of the neck, and truncated cones for the upper arm, which won't account for the shoulder. The foot is misrepresented as the shape only accounts for the mass between the ankle and toes. Overall, considering the large number of required anthropometric measurements and inaccuracy in representing complex body shapes, geometric modelling is an inappropriate method for determining subject-specific BSPs. The scanning method presented in this thesis is a more accurate and faster alternative.

5.5.6 Group-specific regression tables

The male groups have a more significant requirement for group-specific regression tables than the female groups, with young males showing more significant decreases in NRMSE than any other group. The inertial properties showed the largest sensitivity to group-specific regression. The majority of groups featured in this study presented with significantly lower NRMSE values than the All group. The BMI-based groupings only showed significantly lower NRMSE values for inertia when compared to the All group. As BMI assesses differences in body mass (although fails to distinguish between muscle and fatty mass), it seems logical that inertia would be greatly affected by smaller or larger segment mass. Despite this, neither BMI-related group show any differences in segment mass NRMSE values.

The size of each group will undoubtedly affect the accuracy of each set of regression calculations. Assuming the BSPs are similar between members of the specified group, larger populations will have regression values more representative of the population. For the groups within this study who presented with significantly smaller NRMSE values, the sizes tended to be approximately half of the All group, or greater than half. Future work may look to find the optimal group size for the most relevant regression table. As previous studies have presented such large differences between the young and elderly, a larger sub-group of elderly participants would have been desirable. However, the current protocols would be unsuitable for participants unable to support themselves and the elderly may have been unsteady on the rotating platform. Iterations of the scanner design should look at addressing these issues.

5.5.7 Limitations

Differences in estimations of the BSPs of the feet and head were not ideal. Further work on how to address field-of-view limitations must improve the representation of these segments. The head is not included in the upper body MSK model featured in this thesis, however future models looking at the spine or neck may warrant further investigation into head BSPs. Simply removing the hair data from the point-cloud may resolve this issue. This would be important in defining the correct inertial properties of the accelerating head segment, which would be of interest when considering head/neck interaction. Volume data relating to the neck was removed in order to find BSPs for the head on its own, however, this portion of the body would also be of interest when looking at the movements of the spine or head. Minor differences between the literature and the upper arm highlights the need to separate the shoulder from the arm data. This would be of interest for those using shoulder MSK models (presented in Chapter 4).

All segments were defined based on externally-visible anatomical landmarks, barring the precise location of the knee joint and proximal points of the thigh, the latter of which would ideally be based on a plane through the groin and ASIS. The former was taken as the most anterior (and visible) point of the patella which forms a clear apex around the knee joint. As the patella is generally more superiorly placed than the tibial-femoral joint, this may have impacted on the shank CoM however the true effects are unclear. Subjects' right and left ASIS points were not visible landmarks on the scanning output data, which could only be rectified by manual palpation prior to the scan. The alternative, estimating the hip joint centre, was not entirely based on regression. Instead the ratio of measured to expected shank length was applied to the expected thigh length as per De Leva's (1996) regression tables. Nonetheless, the partial use of regression data means the specificity of the thigh is diminished compared to the other, manually split segments. Future protocols may look to include a measurement of the right and left ASIS positions, relative to the global coordinate frame, for each subject. This should also be considered for the knee.

5.6 Conclusions

This chapter has presented a new method for calculating subject-specific body segment parameters with an accuracy in line with empirical data from cadaver studies and geometric modelling. Such methods can easily be integrated into a clinical or research setting, providing users with rapid access to subject-specific BSPs or the potential to create population-specific regression tables based on groups of interest. Both the comparisons of absolute data with the literature, and subject-specific data with geometric modelling, serve as confirmation that the BSPs calculated from the 3D scanning method are accurate to the point at which they will not significantly affect the outputs from MSK modelling. The BSPs were found to be well within the ranges found in the literature, however differences existed due to limitations of the scanner and natural variations between the group scanned in this study and other studies. For example this study grouped both males and females, whilst Durkin & Dowling (2003) show clear differences between sexes. The majority of cadaveric studies only present male data, and as shown in Section 5.4.6 removing female data from the current study reduced the differences between measured and regression-estimated values for the male groups.

There is certainly scope to improve regression tables based on specific groupings. Due to errors in existing regression-produced BSPs of anywhere up to 40% between studies, it seems sensible to consider group-specific regression tables. It was found that young males appeared to benefit more than any other group from specific regression tables, and with larger numbers this could possibly be confirmed for other populations too. This would suggest grouping by both gender and age could improve regression table estimation. Regression calculation of inertial characteristics was significantly improved by various groupings: age, gender and BMI. Considering the speed with which populations can be scanned, it would therefore be recommended to group specifically using all three of these traits depending on the aims of the study (see earlier discussions on the need for subject-specific inertial properties). The groupings chosen, and the optimal base number of scanned subjects, need further investigation. However, the rapidity of the presented 3D scanning method makes this possible.

Avenues for future work include the application of subject-specific BSPs to the work presented in the remaining chapters of this thesis. The discussions in Section 5.1

suggest that bespoke BSPs would only be necessary for certain activities, however, as the methods proposed in this chapter present a rapid method for BSP calculation it would be sensible to consider subject-specific data for any study. The equipment is inexpensive when compared to the previous medical imaging methods, and the processing times, when included in MSK modelling requirements, would be negligible. Existing uses for 3D scanning technology includes bespoke tailoring, and in a similar way could be used to design custom-made prosthetics or orthotics. Exoskeletons, as introduced as Chapter 4, would be one such example where 3D scanning could be used to design the components integral to a device's success for individual patients.

An additional suggestion for future work to improve processing times would be to develop automatic segmentation of the body scan, this has been shown to have varying success by previous authors (Leong et al., 2007; Zhiqiang et al., 2012; Sheets et al., 2010; Zhong & Xu, 2006). Developing this would need a thorough validation, made possible with the data from the current study which produced anatomical landmarks for all subjects.

Despite the success of the presented protocols, subject-specific BSPs will not be used for the main study in this thesis. This is due to two main reasons:

1. The 3D scanner used was on temporary loan from an external company and was returned before the motion capture study was begun, and;
2. In its current form, the 3D scanning product would not be suitable for an elderly group due to instability in the rotating platform.

However, the data presented in this thesis can be used as alternative regression tables, based on living subjects and validated against empirical, cadaveric studies. Not using bespoke BSPs will not have a significant effect on the outcomes of the MSK modelling work of this thesis. As discussed previously, MSK models are most sensitive to BSP errors when the observed motions have high accelerations and include no additional external forces (not considering gravity). The sit-to-stand motion will be performed at a self-selected speed in all groups, and there are additional external forces at the feet, seat

and hands. Therefore, not using the 3D scanner will not be detrimental to the work presented.

The variations in segment mass between age groups and sexes is well known, and the method presented here could be used to determine anthropometrics for a number of different populations. This is important when creating bespoke inputs for MSK modelling, however, it would also be sensible to consider differences in strength and susceptibility to muscle weakness when building these subject-specific models. Chapter 6 considers the theoretical effects of muscle weakness on young kinematics and the MSK loadsharing optimisations. This furthers the investigation into catering MSK models to different age groups, and could provide ways by which to customise all future simulations.

CHAPTER 6

MODELLING MUSCLE WEAKNESS DURING A SIT-TO-STAND ACTIVITY

This chapter focuses on the theoretical effects of local muscle weakness on healthy kinematic data during a sit-to-stand activity by testing the ability of musculoskeletal models to solve the natural muscle loadsharing redundancy with a reduced capacity for muscles to produce force.

6.1 Introduction

MSK models rely on the calculation or estimation of muscle force upper boundaries, as discussed in Chapter 4, to simulate a physiological strength profile. Such parameters should be made adaptable to patients who have experienced a significant reduction in muscle mass or volume, resulting in a loss of muscle strength for reasons including: amputation (Sherk et al., 2010; 2008), disuse (Wisdom et al., 2014) or ageing (Frontera et al., 2000; Lexell et al., 1988; Young et al., 1985).

The effects of theoretical muscle weakness have been comprehensively explored (Cleather & Bull, 2012; Bull et al., 2008; Monaco & Micera, 2012; Carmichael & Liu, 2013; Afschrift et al., 2014; van der Krogt et al., 2012; Barrett & Lichtwark, 2008; Thelen, 2003; van Drongelen et al., 2006; Komura et al., 2004). These studies implemented a number of different approaches to simulate both local and general weakness, but most are based on arbitrary scalar reductions to existing maximum force boundaries. A summary of these can be found in Table 6.1 , where studies utilising well-established MSK models are reviewed, together with the clinical applications.

The majority of studies in the literature focus on the effects of lower limb muscle weakness and therefore the upper limb is poorly represented. The arms and shoulders are vital in carrying out many activities of daily living (ADLs), particularly for the elderly who rely heavily on their upper extremities for maintaining independence. The studies which did focus on upper limb weakness only simulated general muscle weakness and, unlike the lower limb studies, did not address the effects of localised atrophy. Only van Drongelen et al. (2006) have published a review of muscle weakness in the upper limb as quantified through arbitrary inhibition of muscle activation based on anatomy; this shows some correlation with spinal cord injury data, depending on the lesion level chosen. This does not feature clinically applicable ADLs but can provide relevant information on tertraplegics, due to known information on local muscle paralysis.

Table 6.1 - Studies simulating muscle weakness (MW). LL = lower limb, UL = upper limb. Software/MSK model used : [1] - Generic MSK models in Opensim (Delp et al., 2007), [2] - UKNSM (Charlton & Johnson, 2006) and FreeBody (Cleather & Bull, 2015), [3] - Delp et al. (1990), [4] – Delft Shoulder and Elbow model (DSEM - van Der Helm, 1997), [5] - Holzbaur et al. (2005)

Studies	Assessment of the effects of MW	Outcomes
Afschrift et al., 2014	Tolerance of measured kinematics of 3 activities to MW: Quantified as the presence of a difference between required joint torques imposed by task, and moment generation by a muscle model, termed the capability gap (CG). Muscle force upper bounds reduced for all LL muscles[1]	Reducing the maximum muscle force by 30% led to a CG during stair ascent/descent, and 40% for gait. Sit-to-stand most tolerant, with up to 70/80% weakness before a CG was found
Carmichael & Liu, 2013	Matrix of impairments (of various severities) based on stroke patient data, applied to strength estimation model[5]. Assessed difference in CG, as above, when drinking from a bottle	Not all impairment severities hindered user capability, the CG between subject strength and task requirement varied throughout motion. MW is not always inhibiting
van der Krogt et al., 2012	Assessment of the tolerance levels of gait to MW. All LL muscles weakened, then as groups, in 20% steps. Tolerance quantified by changes in joint angles, use of reserve muscles or an unsuccessful muscle force loadsharing optimisation[1]	Up to 40% MW tolerated. Compensations could cause further overuse issues through increased antagonistic action
Bull et al., 2008	Inhibiting muscle activation during various UL and LL ADLs[2]	Tasks theoretically unachievable in weakened state
van Drongelen et al., 2006	Investigation into the relationship between MW and shoulder joint loading during weight lifting. MW of UL muscles related to a lesion level at known innervation points on spine[4]. Success in loadsharing optimisation in FD model also assessed	When over 80% of muscles had a reduced maximum force, no solutions were found. At 45%, simulations were possible. Minor effects seen on GH contact force. Shows how kinematics can be altered at various weakness tolerance levels in order to complete task
Komura et al., 2004	Full deactivation of individual LL muscles to observe effects on other muscles and joint contact forces during gait[3]	Vital muscle paralysation did not stop motion, made possible by synergy re-arrangement. Rehabilitation should train alternative muscles. Some effects were seen on joint contact forces despite a lack of affiliation to deactivated muscles

There is a lack of evidence to support the arbitrary weakening of individual muscles or groups in computational modelling studies, resulting in poor clinical outcomes. A comprehensive analysis on the effects of independent muscle weakness would give an understanding of the key muscles required for movement. Studies in the literature focus on weakened joint power (Afschrift et al., 2014) or effects on joint contact force (van Drongelen et al., 2006) and so do not enable this identification of key muscles. van der Krogt et al. (2012) presented a full review on the robustness of human gait to muscle weakness. They simulated arbitrary muscle weakness in a forward dynamics musculoskeletal model through incremental 20% reductions of the maximum muscle force. This was used to quantify two effects: tolerance to weakness (through loadsharing success or changes in kinematics, made possible through FD), and the increases/decreases in neighbouring muscle activations. For a generalised tolerance, where all muscle force upper bounds were lowered, it was found that up to 40% weakness could be tolerated before a mathematical solution could not be found or a major deviation from baseline movements was observed.

It is important to note that the kinematic data from which the tolerance levels were simulated in van der Krogt's study came from adolescent subjects. This may change for kinematics observed from older subjects; the movements employed to successfully complete an activity can change significantly with age-related alterations (Chapter 3). Furthermore, many studies do not state how the force upper boundary of individual muscles is defined, a key variable in such work. Studies which quantify, and account for, subject strength are based on complex models of contraction dynamics, which show large differences when compared to published strength data (Carmichael & Liu, 2013). Therefore, it cannot be definitively stated that the tolerances to muscle weakness presented by these studies are purely a result of the MSK modelling changes employed. It may be the case that, in reality, the human body can tolerate much higher levels of weakness through increased activation of remaining musculature, but the modelling studies did not use physiologically correct maximum muscle force values.

One of the main disadvantages of modelling theoretical muscle weakness is an inability to validate such work. Work in the literature demonstrates a number of ways to simulate feasible levels of muscle weakening, and are based on reductions in muscle mass (Monaco & Micera, 2012), strength (Thelen, 2003) or a combination of all age-

related changes (Barrett & Lichtwark, 2008). However, these may not accurately reflect clinical muscle weakness which is realistically affected by changes in muscle architecture and neuromuscular excitation.

One MSK modelling approach would be to have bespoke strength profiles for individual muscles, detailing how much tension each can produce to correctly inform the mathematical optimisation and solve for accurate muscle loadsharing. This is impractical due to the large number of properties required for such models. A forward dynamics (FD) MSK model features such properties by implementing a Hill type muscle model (Hill, 1938; Winters, 1990) which takes into account the detailed effects of muscle contraction dynamics such as neural-excitation, activation and force-length-velocity relationships (Prinold, 2012). The Hill type muscle model is of most relevance when implemented in FD-MSK models; it allows the calculation of tendon-path excursion in calculating joint kinematics. It is not used in this thesis as both featured models will use ID. However, other studies have shown its ability to simulate ageing muscle (Thelen, 2003). The work by Thelen simulated aging contraction dynamics for use in forward dynamic MSK models. Literature values chosen by Thelen produced a more relevant model of muscle weakness through the comprehensive use of empirical data to formulate the correct contraction dynamics of an older body, improving on the previously arbitrary choice of strength profile. However, the literature is rarely consistent when defining the variables required to produce accurate contraction mechanics, and the mean values used in that study may not reflect subject or muscle-specific weakness (Hasson & Caldwell, 2012).

One possible approach to selecting muscles at risk of weakness with age is through identifying muscle fibre–specific atrophy. As discussed in Chapter 2, muscle fibre type distribution changes with age, with a proportional increase in Type 1 muscle fibres (slow twitch, low fatigue) and a reduction in the number and size of Type 2 fibres (fast twitch, high fatigue). This would drastically affect muscle contractile properties, particularly muscles with the latter fibres. A summary of fibre type distributions across the muscles in and around the shoulder, as featured in the UKNSM, is shown in Table 6.2. All studies reviewed presented either muscle biopsies on living male subjects or autopsies of male cadavers less than 24 hours after death. Only male-based studies are chosen due to significant differences in fibre type distribution between males and

females (Paoli et al., 2013; Brooke & Engel, 1969; Manta et al., 1996), making grouping unwise and owing to the sole use of male data within this thesis. Muscles with multiple lines of action were grouped together and, due to a lack of human data for the serratus anterior and brachialis, studies on rats (Potluri et al., 2006) and monkeys (Roy et al., 1984) have been included as these were the sole studies found for these muscles. Where appropriate, segregation of Type 2 muscles into further groupings (Type 2a and b) have been summed together and, if reported, deep and superficial samples of the same muscle have been averaged.

Table 6.2 - Fibre type distributions for muscles featured in the UKNSM. 1 - Srinivasan et al., 2007, 2 - Johnson et al., 1973, 3 - Potluri et al., 2006, 4 - Roy et al., 1984, 5 - Le Bozec & Maton, 1987, 6 - Elder et al., 1982, 7 - Nygaard et al., 1983, 8 - Mavidis et al., 2007, 9 - Baker & Hardy, 1989, 10 - MacDougall et al., 1984, 11 - Dahmane et al., 2005, 12 - Kadi et al., 1998, 13 - Paoli et al., 2013, 14 - Kronberg et al., 1997, 15 - Mandroukas et al., 2010, 16 - Lovering & Russ, 2008 and 17 - Manta et al., 1996. No data could be found for the Levator Scapulae (LS) or Pronator/Supinator Humerus (PT_hum, SUP_hum)

Muscle (abbr.)	Mean Type 2 fibre proportion (% total number of fibres)	Weighted average
Trapezius (Trap_scap, Trap_clav)	46[2], 36[12]	39
Rhomboids (Rmaj, Rmin)	55[2]	55
Serratus Anterior (SA)	82[3]	82
Pectoralis minor (Pmin)	65[1]	65
Pectoralis major (Pmaj_clav, Pmaj_thor)	65[1], 58[2] (Clavicle), 57[2] (thoracic)	59
Latissimus Dorsi (LD)	52[1], 49[2], 47[9], 74[13]	57
Deltoid anterior (DeltA)	52[1], 58[2], 53[14], 42[17]	49
Deltoid middle (DeltM)	53[1], 58[2], 47[8], 56[14], 57[15], 42[17]	48
Deltoid posterior (DeltP)	44[1], 58[2], 42[17]	47
Supraspinatus (SS)	50[1], 41[2], 46[16]	45
Infraspinatus (IS)	52[1], 55[2], 59[16]	56
Subscapularis (SBS)	63[1], 62[16]	62
Teres minor (Tmin)	51[1], 51[16]	51
Teres major (Tmaj)	52[1], 60[16]	57
Coracobrachialis (CB)	77[1]	77
Biceps short (BIC_short)	63[1], 46[2], 55[6], 58[7], 61[10], 53[11], 52[17]	57
Biceps long (BIC_long)	61[1], 46[2], 55[6], 58[7], 61[10], 53[11], 52[17]	56
Triceps long (TRI_long)	57[1], 67[2], 64[5], 60[6], 62[11]	62
Triceps medial (TRI_med)	20[2], 40[6], 62[11]	48
Triceps lateral (TRI_lat)	67[2], 60[6], 62[11]	63
Brachialis (Bra)	32[4]	32
Anconeous (Anc)	33[5]	33
Brachioradialis (BRD)	60[2], 50[11]	53

Most muscles contain an equal balance of Type 1 and 2 muscle fibres, however muscles such as the serratus anterior, pectoralis, subscapularis, coracobrachialis and two

heads of triceps contain over 60% Type 2. This may suggest an increased susceptibility to weakening with age due to the associated loss of Type 2 fibres, and further investigation into the specific weakness of these muscles could potentially explain difficulties in ADLs. Function of the serratus anterior muscle during standing and sitting, for example, has not previously been investigated. Considering its role in stabilising the scapula during elevation, weakness of this muscle may prove detrimental to the success of an MSK model in producing a solution to the loadsharing optimisation. The pectoralis muscles are important in arm flexion at the shoulder, a movement common in many ADLs, and again weakness of such muscles may hinder everyday life. In many cases, the studies reviewed did not specify internal muscle lines of action, therefore the deltoids have sometimes been grouped as one muscle. This also applies to the biceps and triceps as shown by repeating values across muscle heads. In terms of averaging superficial and deep values, in some cases these were found to be statistically significantly different and may prove more or less susceptible to ageing if the depths of the muscle fibre samples were specified.

Sit-to-stand (STS) is a motion noticeably missing from the literature in the context of simulating upper limb muscle weakness. It is well established that diminishing lower limb strength negatively affects STS performance (Hughes et al., 1996; Bohannon, 2007; Bohannon, 2009; Gross et al., 1998). However, the effects of weakening the upper limb muscles have not yet been investigated. This is of particular interest in any study on ageing, considering the importance of using arm rests in achieving a successful STS motion (Chapter 3). Due to the loads placed on the shoulder during STS with arm rests (Anglin et al., 2000), it is assumed that the scapula would play an important role in stabilising the upper body during the motion. This in turn would suggest weakness of the scapula stabilisers, such as serratus anterior, the rhomboids and trapezius (Paine & Voight, 1993), would be the most sensitive determinants of STS success. Existing palsy conditions such as scapula winging (Martin & Fish, 2008), although not specifically related to ageing, is an example of serratus anterior weakening. Confirmation that atrophy of these muscles is a common trait, and understanding the effects of this form of weakness, may help to better understand the potential effects of age-related weakness.

The remaining sections of this chapter document an investigation into the theoretical effects of STS performance when upper limb muscles experience extreme weakness. It is hypothesised that weakness of scapula stabilisers will be a major determinant in STS success, and that effectively removing these from a MSK model will hinder its ability to solve the loadsharing optimisation. The main objectives of this study are exploratory, in addition to identifying the vulnerability of scapula stabilisers. Few studies have investigated specific weakness of upper limb muscles in basic activities. The proposed methods described below will therefore be tailored towards specific muscle injury as opposed to generalised weakness. The aim will be to identify how these localised weaknesses affect an ADL.

6.2 Methods

Muscle weakness was simulated using the UKNSM (Chapter 4). 27 healthy male subjects (aged 28 ± 5 years) were asked to perform a STS activity three times, ensuring use of arm rests. Applying an external force to the upper limbs ensured higher loads through the shoulders and a better understanding of vulnerability to muscle weakness. Retro-reflective markers were placed on the hands, arms and torso of each participant using hypoallergenic tape and data was captured using a 10-camera motion capture system (Vicon Motion Systems, Oxford, UK). External force data was obtained through load cells placed in the arm rests of an instrumented chair (Duffell et al., 2013) with a set seat height of 46 cm. Two single point load cells were used to measure vertical hand forces (Vishay, model 1042-0050-G506R) and two tension/compression load cells were used for the measurement of anterior-posterior hand forces (Omega, model LCM703-25) as shown in Figure 6.1. Ethical approval was received through the Joint Research Compliance Office and Imperial College Research Ethics Committee. The full application and approval letter can be found in Appendix 3.

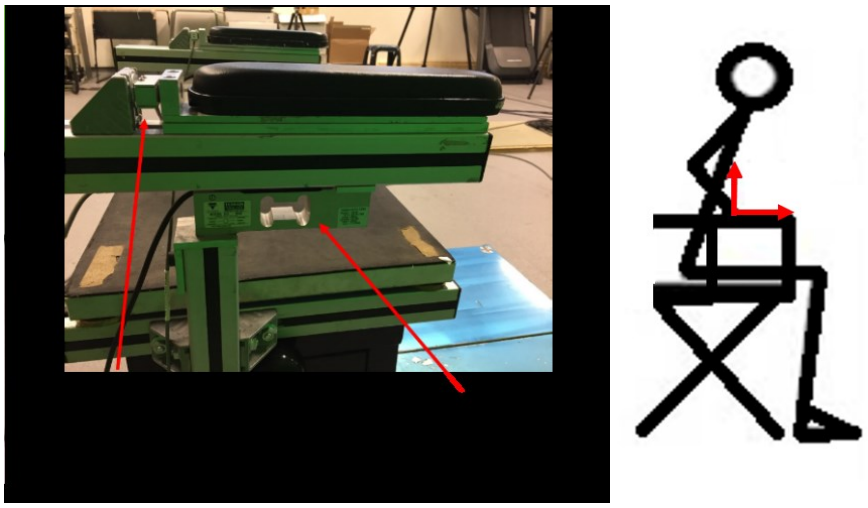


Figure 6.1 - Instrumented arm rest. A positive force is defined as being in the superior (single point load cell) and anterior (tension/compression load cell) directions. The force of a participant pushing down on the arm rests is measured, as visualised in the line drawing, where the red arrows represents the reaction force components.

Initially, the UKNSM was used to obtain muscle and joint contact forces for the non-weakened state. As discussed in Section 6.1, the difficulty in obtaining subject-specific muscle force upper bounds was addressed by effectively creating an unbounded optimisation. This was previously used by Prinold (2012) as a method to identify pivotal muscles in extreme activities, and in this study will be used in much the same way: identifying the muscles which increase their activity to compensate for other muscles' weakness.

The scenario of extreme muscle weakness was simulated by independently removing each of the 29 muscles modelled within the UKNSM, effectively leaving the motion with only 28 functional muscle lines of action for each repetition of the analysis. STS performance was evaluated based on the number of successful loadsharing solutions produced by the loadsharing optimisation. Activation of the remaining muscles and changes in glenohumeral and elbow joint contact force were observed for motions producing a successful loadsharing solution.

6.2.1 Data analysis and statistics

Failed optimisations were initially calculated on a trial-by-trial basis. For each participant, the number of time frames with failed optimisations were found for each muscle disabled for each of the three trials. These time frames were converted into a percentage of the trial length (rounded to the nearest integer), and if a failure occurred at that percentage for at least two of the trials, it was included in the subject's failure count. The number of time frames with failures was then summed for each participant.

Related-sample sign tests were used to test the hypothesis that there is a difference in peak muscle (and summation of all peak muscle forces) and joint forces between full strength and weakened-muscle scenarios. This test evaluates the significance of the median difference between the full strength and weakened states. The number of positive and negative differences between the two related samples are considered – if the samples are very similar the expected median difference would be close to zero. A standard threshold of $p < 0.05$ is set to determine whether the effects seen are statistically significant or could have occurred through chance. If the calculated p-value is lower than this threshold, then the null hypothesis is rejected and a significant increase or decrease in muscle/joint reaction force has been found under the weakening of a particular muscle.

6.3 Results

The prevalence of failed optimisations are shown in Figure 6.2. The serratus anterior proved the most debilitating, with the UKNSM unable to produce solutions during portions of the motion for 18 of the 27 participants. Others were found to have no solutions for weakness of the clavicular portion of the pectoralis major (2 participants with 33% and 40% of the motion failed), posterior deltoid (1 participant with 34%) and long head of biceps (2 participants with 5% and 21%). All other participants had 100% successful optimisation solutions.

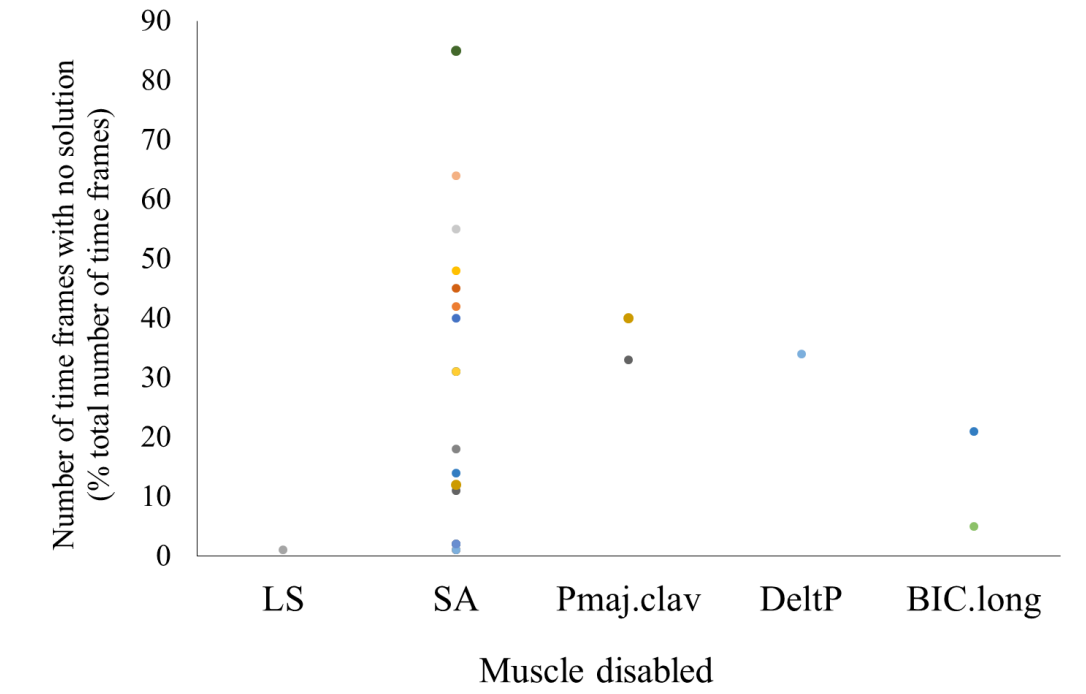


Figure 6.2 – Mean number of failed time frames for all participants whose data included failed optimisations. Colours represent different participants

Tables 6.3, 6.4, 6.5 and 6.6 show the changes in peak muscle and joint force. Each muscle removed resulted in significant increases and/or decreases in peak force in at least one other muscle ($p<0.0005$ to $p = 0.043$). 14 resulted in significant changes in glenohumeral force ($p<0.0005$ to $p = 0.021$) and 8 similarly for the elbow ($p<0.0005$ to $p = 0.021$). The serratus anterior muscle caused the most extreme changes. 18 out of the 28 remaining muscles had significantly increased peak muscle forces ($p<0.0005$ to $p = 0.021$), and 4 had decreased peak muscle forces ($p<0.0005$ to $p = 0.007$) when serratus anterior was disabled. This also resulted in large statistically significant increases in both the glenohumeral and elbow resultant contact forces (Table 6.6; $p<0.0005$).

The following tables and figures include data for the SA muscle which can be considered non-physiological owing to the unbounded nature of the loadsharing optimisation. These have either been highlighted in red in tables or placed on separate axes for graphs. These data have been included to emphasise the effects of this muscle's absence, but should not be considered as factual forces as it is likely that in these cases the task would not be able to be completed and the subjects would most likely have to alter the kinematics as well as the kinetics of the task.

Table 6.3 - Peak muscle forces of each muscle modelled in the UKNSM for a full-strength model and subsequent weakening of individual muscles as listed in the first column. * ($p<0.05$) or ** ($p<0.0005$) show that removing the muscle in this row caused a statistically significant difference in peak muscle force for the muscle in this column when compared to the full-strength model

	Median peak muscle force (N)							
Muscle weakened	Anc	BIC_long	BIC_short	Bra	BRD	CB	DeltA	DeltM
Full strength	**42.8**	**226.5**	**11.6**	**40.2**	**9.8**	**1.1**	**174.9**	**256.4**
Anc		225.4	11.5	40.2	9.8	1.0	175.9**	258.4*
BIC_long	42.0**		106**	62.5**	14.5**	0.3	193.6**	645.4*
BIC_shor	42.8	226.6*		47.7	9.8*	2.1*	174.9	256.4
Bra	42.8	226.6	20.8*		19.4**	1.1	174.9	256.4*
BRD	42.8	226.6	14.1*	48.2**		1.1	174.9	256.4
CB	42.8	226.5	12.9*	40.2	9.8		174.9	256.4
DeltA	70.3**	322.5**	11.4	39.8*	9.8*	0.8		431.6**
DeltM	55.3**	240.2	5.7**	35.1	6.3	0.9	192.7*	
DeltP	42.8	226.6	11.6	56.2**	11.6**	1.1	174.9	442.6**
IS	42.8*	226.6	11.9*	41.1**	9.8**	1.1*	174.8	256.4
LD	42.8*	227	15.4*	47.6*	10*	1.3	174.7	270.4
LS	43.4**	226.5	11.5	40.2	9.8	1.0	175**	263.0*
Pmaj_clav	53.2	286.0	20**	45.9	9.9	1.6*	116.7	399.4
Pmaj_thor	69.0*	516.6*	71.8**	38.8*	7.5	47.7**	260.1**	242.2*
Pmin	42.8	227.5	11.6*	46.4*	10	1.1*	174.3	256.3*
PT_hum	42.8**	226.6	11.6	40.3**	9.8**	1.1	174.9**	256.4*
Rmaj	43.4*	230.4	11.7*	42.8	9.8	1.4*	175.3	256.7
Rmin	43.1*	227.2	11.6	40.3	9.8	1.1	175.1*	258.5
SA	225035.9**	847710**	89374.8	183036.6*	130366.6**	274789.1**	334.2	1108622**
SBS	42.8	226.5	14.3**	40.2	9.6	1.1*	178.2	256.4*
SS	59.2**	267.6*	7.1**	39.5*	8.2**	0.4	189.7**	266
SUP_hum	42.8*	225.7	11.6*	40.2**	9.8**	1.1	174.9**	256.4*
Tmaj	67.4**	303.4*	10	42.8	9.1	0.8	184.7	238
Tmin	42.8	226.6	11.6*	40.2	9.8	1.1	174.9	256.4
Trap_cla	43.9*	227.2	11.6	43.1*	9.5*	1.1	175.5**	256.9
Trap_sca	48.7*	265.9**	18.4**	50.3	11.1	2.3**	177.9	265.6
TRI_lat	46.5**	208.7	10.1**	40.2	9.8	0.7	175.4*	258
TRI_long	52.0**	195.3*	11.1	40.2	9.8	1.8**	173.8	256.9
TRI_med	51.5**	207.6	10.2**	38.3*	7.4*	1.0	178.6**	271.4*

Table 6.4 - Peak muscle forces for each muscle modelled in the UKNSM for a full-strength model and subsequent weakening of individual muscles as listed in the first column. * ($p<0.05$) or ** ($p<0.0005$) show that removing the muscle in this row caused a statistically significant difference in peak muscle force for the muscle in this column when compared to the full-strength model

	Median peak muscle force (N)							
Muscle weakened	DeltP	IS	LD	LS	Pmaj_clav	Pmaj_thor	Pmin	PT_hum
Full strength	**132.1**	**17.2**	**22.8**	**50.0**	**45.8**	**443.4**	**23.7**	**1.3**
Anc	132.1*	17.2	22.8**	52.5**	49.7	444.2	23.6**	1.4*
BIC_long	155.3	6.9**	38.4**	50.0	55.3	720.7*	44.2**	1.5*
BIC_shor	132.1	18.4*	23.2	50.0	45.8	443.4	23.7	1.3*
Bra	133.5*	19.7**	23.4**	50.0	45.8	443.4	24.2*	1.6*
BRD	132.3*	18*	22.9	50.0	45.8	443.4	23.7*	1.3*
CB	132.1	17.2	22.8	50.0	45.8	444.0	23.7	1.3
DeltA	132.1	15.7	21.3	85.3**	80.1**	535.3**	23.6	2.3**
DeltM	271.9**	19.2	36.7**	72.3**	56.0**	435.2	12.6**	2.0*
DeltP		66.4**	28.2*	49.7	48.1	443.4	33.3*	1.3
IS	132.1		22.8	50.0	45.8	443.4	23.7	1.3
LD	132.1	16.7		48.8	45.8	443.4*	24.3**	1.3
LS	132.4	17.2	22.6		52.7*	442.0	23.6**	1.4*
Pmaj_clav	136.5	18.4*	20.7	84.7		503.7*	27.7	2.0
Pmaj_thor	131.2	25.8	181.2**	88.0**	167.4**		27.7	1.9*
Pmin	136.3**	17.2	23.3*	51.2	44.7	443.4		1.3
PT_hum	132.1	17.3*	22.8	50.0**	45.8*	443.4	23.7	
Rmaj	133.2*	17.2	27.3	54.5**	47.0*	443.5*	25.4*	1.4*
Rmin	132.3	17.2	23.4	53.9**	47.6*	443.0*	23.3	1.3*
SA	240822.5**	181065.3*	0.7**	64.0	235.6	542.0*	645.0**	85402.1**
SBS	132.1	34.3*	22.9	50.0	45.8	443.4	23.6	1.3
SS	136.5	32.2**	19.2	57.4*	84.8	451.2	23.5	1.6
SUP_hum	132.1	17.2*	22.8	50.0**	45.9	443.4	23.7*	1.3**
Tmaj	268.0**	15.2	191.7**	93.5**	82.9	382.7*	14.8**	1.9*
Tmin	132.1	17.3	22.8	50.0	45.8	443.4	23.7	1.3
Trap_cla	133.2*	18.1*	17.9**	51.0	40.6	441.3*	20.8**	1.3*
Trap_sca	137.5	18.3	19.1	49.8	70.4	418.5	22.4	1.6*
TRI_lat	132.1	17.2	25.9**	49.1*	46.5	449.7	23.7**	1.5**
TRI_long	133.2*	17.2	19.2*	57.4**	56.2*	440.3*	23.6	1.4
TRI_med	131.0*	17.1	25.8**	50.8	57.5	453.5	29.8**	1.7*

Table 6.5 - Peak muscle forces of each muscle modelled in the UKNSM for a full-strength model and subsequent weakening of individual muscles as listed in the first column. * ($p<0.05$) or ** ($p<0.0005$) show that removing the muscle in this row caused a statistically significant difference in peak muscle force for the muscle in this column when compared to the full-strength model

	Median peak muscle force (N)							
Muscle weakened	Rmaj	Rmin	SA	SBS	SS	SUP_hum	Tmaj	Tmin
Full strength	**48.0**	**22.4**	**382.8**	**21.9**	**166.2**	**3.6**	**209.4**	**0.8**
Anc	48.0	22.9**	397.8**	21.8	167.4**	3.8**	212.6*	0.8
BIC_long	105.6*	36.0	730.0	7.7**	680.1**	3.6	483.7**	0*
BIC_shor	48.0	22.4*	382.8	23.4**	166.2	3.6	209.4	0.7*
Bra	48.0	22.6	382.8	22.6*	166.2*	3.6*	209.4	1.0*
BRD	48.0	22.4	382.8	22.1*	166.2	3.6	209.4	0.8
CB	48.0	22.4	382.8	22.0*	166.2	3.6	209.4	0.8
DeltA	51.7*	35.8*	743.1**	17.2	184.4	6.7**	228.9*	0.5
DeltM	48.7	33.9**	538.4**	16.8*	180.3*	5.1*	146.1**	0*
DeltP	35.1	22.5	382.8	18.7	166.2	3.6	334.5**	23.6**
IS	48.0	22.4	382.8*	37.9**	166.2	3.6	209.4	1.3**
LD	49.0*	22.4	382.8	23.0	166.2	3.6	213.0*	0.8
LS	52.0**	27.0**	393.9**	21.7	166.7*	3.9**	212.0*	0.7
Pmaj_clav	50.4	35.2	679.2	25.7	202.6	4.5*	209.4	0.4
Pmaj_thor	38.5	37.8	776.8**	51.9**	312.7*	4.4	194.5	0
Pmin	49.5	22.3	383.7*	21.9	166.2	3.6	209.5	0.8
PT_hum	48.0*	22.4**	382.8**	21.9	166.2*	3.6**	209.4	0.8
Rmaj		27.2**	381.2*	21.9	166.2	3.6*	208.5*	0.8
Rmin	53.1		382.6	21.9	166.5*	3.7**	211.7	0.8
SA	17.8**	16.4		128388.6	643672.0**	13672.8**	445099.7**	10668.5**
SBS	38.7*	22.4	392.9		166.2	3.6	212.7	1.7**
SS	50.4	37.5*	727.0*	52.3**		5.0	196.8	0.6
SUP_hum	48.0*	22.4**	382.8**	21.9	166.2*		209.4	0.8
Tmaj	37.6	38.1*	757.8**	21.1	172.9	4.9**		1.4*
Tmin	48.0	22.4	382.8	22.2	166.2	3.6	209.4	
Trap_cla	48.8**	23.4*	383.4	22.1	168.0	3.6*	210.6**	0.7
Trap_sca	122.7**	50.1**	400.0	23.7	187.5*	4.7*	211.4	0.3
TRI_lat	48.0*	22.4	353.2**	21.7	168.5**	3.8**	213.4	0.8
TRI_long	44.2	22.9*	473.7**	21.9	159.8**	3.6*	235.5	0.8**
TRI_med	49.3	23.5	386.4	21.6**	176.6**	4.4**	214.6	0.7

Table 6.6 - Peak muscle, and joint forces, of each muscle modelled in the UKNSM for a full-strength model and subsequent weakening of individual muscles as listed in the first column. * (p<0.05) or ** (p<0.0005) show that removing the muscle in this row caused a statistically significant difference in peak muscle force for the muscle in this column when compared to the full-strength model

	Median peak muscle/joint reaction force (N)						
Muscle weakened	Trap_clav	Trap_scap	TRI_lat	TRI_long	TRI_med	Glenohumeral joint reaction force	Elbow joint reaction force
Full strength	**51.5**	**223.8**	**172.3**	**108.4**	**234.3**	**886.3**	**729**
Anc	51.5*	228.5*	173.9**	117.9**	244.6**	906.7**	725.8
BIC_long	43.6*	462.7**	140.4	17.2*	217.4	1839.6**	426.4*
BIC_shor	51.5	223.8	172.3	108.4	234.3	886.3	728.9
Bra	51.5	223.8	172.3	108.4	234.3	886.3	728.9
BRD	51.5	223.8	172.3	108.4	234.3	886.3	728.9
CB	51.5	224.5	172.3	108.4	234.3*	886.3	728.9
DeltA	104.8**	244.2*	223.6**	115.5	384.3**	1025.0	949**
DeltM	54.4*	308.5*	183.1	94.6**	311.6*	804.7*	777.8
DeltP	56.5	223.8	171.7	108.4	232.9	928.6*	687.8
IS	51.5	223.8	172.3	108.4	234.3	886.3	728.9
LD	51.5	223.8	172.3	108.4*	234.3	887.1	753.3
LS	54.9*	226.7	163.4**	120.4**	230.9	914.5**	730.2
Pmaj_clav	46.9	267.8	214.6	102.9	300.5	1006.0	864.7
Pmaj_thor	37.8*	209.1	257.6*	260**	353.6*	1309.2*	1387.7**
Pmin	50.6*	218.8	180.8	108.4	234.3	880.5	728.9
PT_hum	51.5*	223.8*	172.3**	108.5**	234.3**	886.4**	728.9*
Rmaj	51.6**	248.3**	186.3	112.1	232.8	886.3	731.9
Rmin	52.8**	235.2**	172.3**	111.5**	233.2	894.1*	730.3
SA	332.6**	516.2*	6.6**	789171.9**	87.2**	1859983.3**	2248828.1**
SBS	54.2	223.8	172.3	98.3	234.3	886.3	728.9
SS	48.8	266.6	217.0*	62.5*	294.4**	813.7	749.1*
SUP_hum	51.5	223.8*	172.3**	108.4	234.3**	886.5*	728.9
Tmaj	78.8**	292.5*	149.0	200.5**	234.3	974.1	805.0*
Tmin	51.5	223.8	172.3	108.4	234.3	886.3	728.9
Trap_cla		260.2*	168.9*	110.6*	232.5	898.7*	728.5
Trap_sca	112.5**		166.4	126.7**	262.4*	957.0*	753.7**
TRI_lat	51.5*	222.6		132.3**	381.0**	891.2	697.1*
TRI_long	56.0**	230.8	188.8**		290.6**	822.7*	589.5*
TRI_med	51.4	232.8**	367.0**	152.3**		944.4**	690.2

Figure 6.3 presents the median sum total of all remaining peak muscle forces for each muscle removed from the UKNSM. The serratus anterior has been placed on a secondary axis due to the relatively large numbers associated with its simulated weakness. Despite the presence of significant differences between full-strength and all other weakened states, only weakness of 7 of the 29 muscles modelled in the UKNSM caused a significant increase in total peak muscle force when compared to the full-strength model (shown by the first profile in Figure 6.3). 13 of the 29 muscles caused a statistically significant decrease in total peak muscle force, however the absolute differences were small (ranging from 0.02% to 1.02% relative to the full strength) compared to the increases in total peak muscle force (1.96% to 54.52%, 158389.56% for SA).

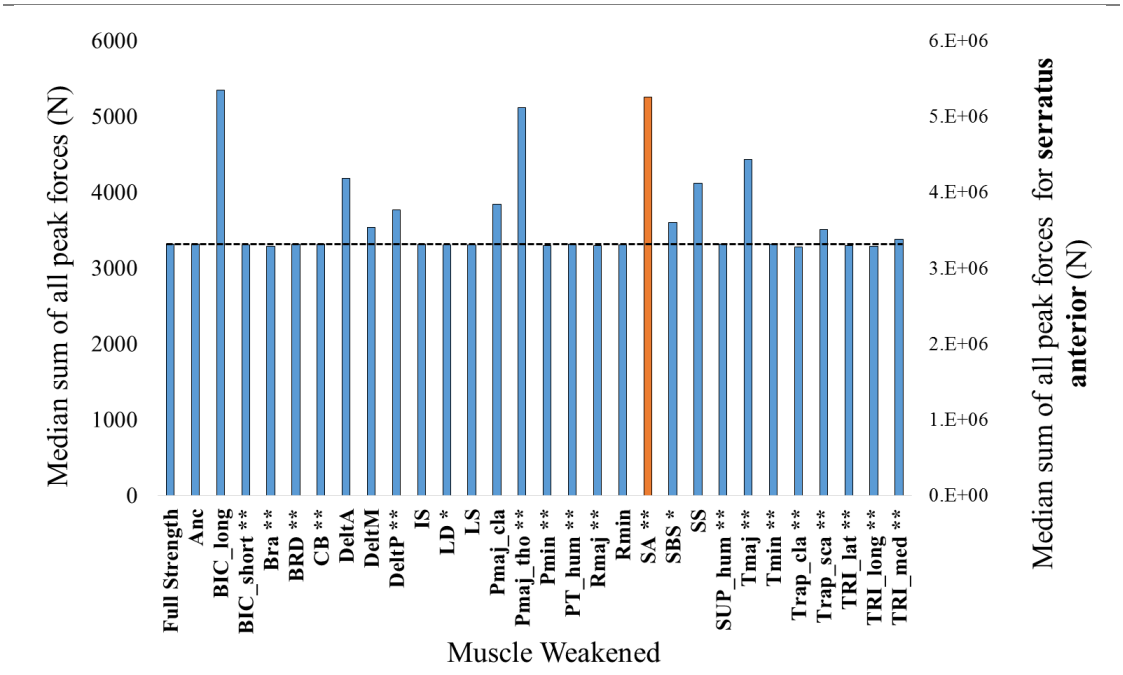


Figure 6.3 – Median sum of all peak muscle forces per weakened muscle. Serratus anterior (orange) has been placed on a secondary axis due to the large forces associated with its removal from the UKNSM. * ($p<0.05$) and ** ($p<0.0005$) indicate statistically significant differences to the full strength state, indicated by the black dashed line and first profile

In addition to the effects of individual muscle weakness on other muscle's peak force activity, the changes in peak joint reaction forces were also observed. Figure 6.4 shows the glenohumeral and elbow JRFs for all muscles whose weakness incited significant changes in peak values. In all but five cases, muscle weaknesses caused

increases in both glenohumeral and elbow JRF, however, weakness of the long head of biceps, middle deltoid and lateral and long heads of triceps resulted in decreases in JRF.

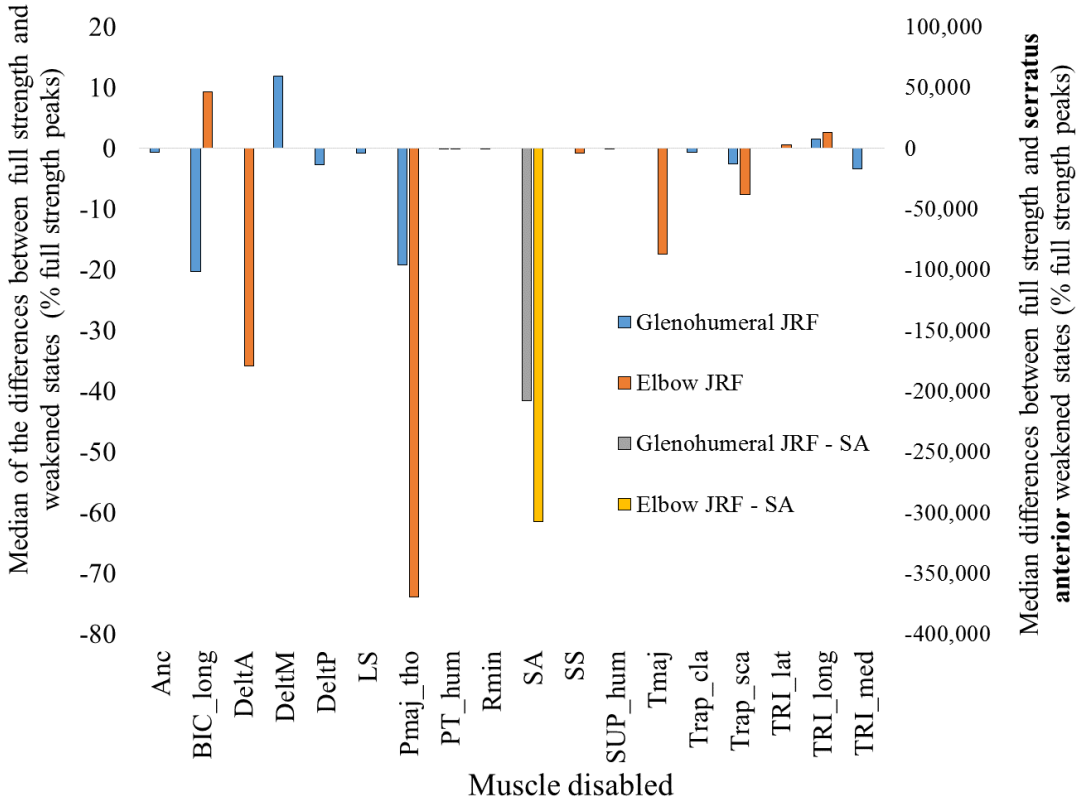


Figure 6.4 - Peak joint reaction force (JRF) changes for the shoulder and elbow. Only muscles whose omission caused a significant change in glenohumeral and/or elbow JRF have been presented. Negative values indicate that the disabled muscles increased the respective JRF. Serratus anterior has been placed on a secondary axis due to the large forces associated with its removal from the UKNSM

6.4 Discussion

The related-samples sign test was used due to non-normal distributions of the data, therefore paired sample t-tests could not be used. Non-parametric tests are generally less powerful than parametric tests and, in the case of sign tests, assess if the median of the differences between two samples is close to zero. As an example, consider the effects of weakening the serratus anterior on the subscapularis. Despite median values of 21.9 N (full strength) versus 128388.6 N (serratus weakened) in subscapularis peak muscle force, no statistically significant difference in median differences was

found. This was due to equal increases and decreases across the 27 participants, with 12 showing increases and 10 presenting with decreases in peak SBS muscle force (the remaining 5 saw no change). Overall, this evidently resulted in a median difference close to zero. The statistical tests do not distinguish the underlying reasons for the differences recorded. There are other possible reasons behind the non-normal distributions of the changes in peak muscle force, including those resulting from a large inter-subject variability. However, the results presented in this chapter show clear effects of disabling certain muscles.

This study has shown that individually disabling all the muscles in turn through the use of MSK modelling resulted in some significant differences in other muscle's activity and joint reaction forces. However, many of these differences, though statistically significant, were very small and are unlikely to be clinically significant. Where relatively large differences were found, peak forces under weakness of other muscles were found to be as high as 300% greater than full strength. Weakness of the serratus anterior stood out as having the largest overall effect on the ability to produce feasible solutions (the UKNSM was unable to solve between 1.0% and 85.0% of the motion); on other muscle's peak activity (overall increase of 158389.6%); and joint reaction force (increase of 209759.3% for glenohumeral and 308381.2% for the elbow). Due to the unbounded optimisation, these peak values are mostly non-physiological, therefore muscle activation (muscle force relative to upper bounds) cannot be calculated. However, this still highlights the detrimental effect weakness of the serratus anterior has on the body during sit-to-stand.

Serratus anterior removal was found to incite changes (mostly increases) in 22 of the 28 remaining muscles in the UKNSM. It was also found to significantly increase both the glenohumeral and elbow resultant joint reaction forces, despite the fact that the serratus anterior crosses neither joint. A similar effect was also seen in the work on muscle weakness during gait by Komura et al. (2004), who found that deactivating the gluteus maximus (which only crosses the hip) affected the forces at the knee and that the forces at the ankle increased when the rectus femoris (which crosses the knee and hip) was disabled. Komura also discovered that weakness of the gluteus maximus affected the knee more than the hip. This can also be seen in the STS data of this thesis where weakness of the long head of biceps brachii increased the glenohumeral joint

force more than the resulting decrease in elbow contact force. The removal of the pectoralis major (which does not cross the elbow joint) also has a major effect on the elbow joint; it almost doubled the contact force. Additionally, and expectantly, weakness in the biceps and triceps caused significant decreases in elbow joint force.

This implies that the effect of localised muscle weakness does not exclusively affect joints which they cross and other muscles with similar lines of action, and it has been shown previously that antagonists of the compensatory muscles will also increase their activity (van der Krogt et al., 2012). This, in turn, affects any joints these secondary or tertiary increases in activation cross. One example of this, from this thesis, is how weakness of the pectoralis major results in changes to elbow JRF. The main effectors of changes in the latter will be the biceps and triceps, both of which were found to significantly increase peak muscle force when the pectoralis major was disabled. Pectoralis major's role in arm adduction is supported by the long head of triceps, which saw an increase of approximately 200% during pectoralis weakness. This action, plus the increased antagonistic action of both the short (590% increase relative to the full strength profile) and long (114%) heads of the biceps brachii, subsequently increased elbow JRF.

The effects shown above come as a result of weakness of the pectoralis major muscle, however, when at full strength the pectoralis as a whole was found to be significantly affected by weakness of the serratus anterior. Relocation of the pectoralis major tendon has been identified as a successful procedure in treating patients with serratus anterior palsy (Streit et al., 2012) and pectoralis minor in many cases is the primary scapula protractor (Castelein et al., 2016), although a shortened length (or increased activation, found in the current study to be a result of serratus anterior weakening) reduces the stability of the scapula (Ludewig & Reynolds, 2009; Cools et al., 2014).

The findings from this study show that both parts of the pectoralis muscle are important during serratus anterior weakness and thus the methods presented could be used in future simulations to inform surgical interventions or predict the effects (primary, secondary or tertiary) of localised weakness. Considering both these findings, and the earlier discussion on the effects of pectoralis major weakness, it also appears as

though the pectoralis is pivotal in the sit-to-stand motion but to a lesser extent than that of serratus anterior.

Weakness of the serratus anterior created significant increases in the peak activity of multiple muscles crossing both the shoulder and elbow. The prime role of the serratus anterior is to stabilise the scapula during elevation, upward rotation and protraction. The synergists for this cover a great deal of the upper limb musculature, including the trapezius, rhomboids, pectoralis minor and levator scapulae. These muscles all have secondary functions which will in turn create tertiary effects on upper limb joints.

It is also interesting to see that serratus anterior is the only muscle to have any noticeable effect on the optimisation success, despite the lack of any boundary on muscle force for synergistic intervention. This would suggest that during certain parts of the motion, the serratus anterior is the only muscle capable of sufficiently stabilising the scapula.

Figure 6.5 shows the prevalence of optimisation failures under serratus anterior weakness. The failures are focused around the start and end of the activity and prove that the serratus anterior is pivotal in the stability of the scapula in more static parts of the motion. For instance, at the start, when participants have their hands placed on the arm rests, the scapula is in an upwardly-rotated and elevated position. Evidently no other muscles can provide the stability here. However, it stands to reason (considering the unrealistically-large forces seen in other muscles while serratus anterior was disabled) that introducing more realistic muscle force upper boundaries for the other muscles may increase the failure rate around the point of peak external hand force. This was found to occur at approximately 35 (± 6.2) % of the trial.

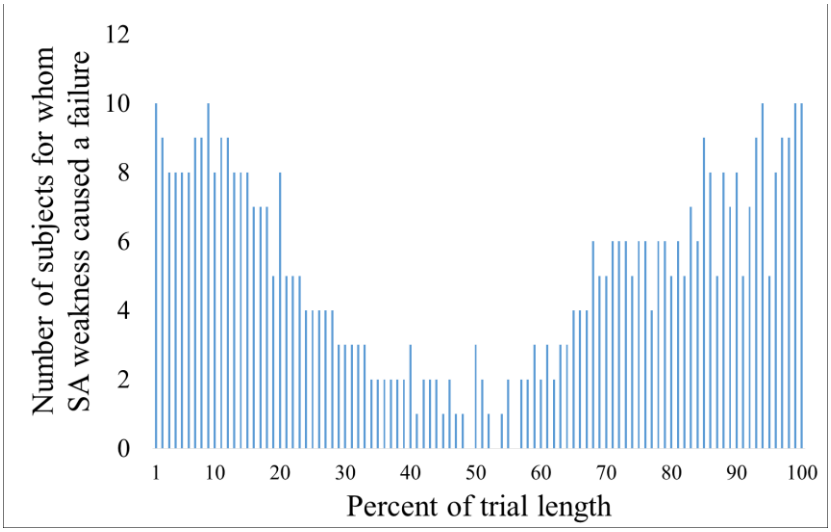


Figure 6.5 - Incidence of failed optimisations for a weakened serratus anterior

Weakness of the serratus anterior is one of the prime causes behind a condition known as 'scapula winging' (Martin & Fish, 2008). Those suffering from scapula winging, caused by serratus anterior palsy, have previously been shown to suffer in terms of pain and functional decline during activities of daily living and sports (Kauppila & Vastamäki, 1996). A review by Cools et al. (2014) found that weakness, or rather reduced muscle activity, of the serratus anterior is commonly associated with shoulder impingement and other pathologies of the upper limb. This shows the clinical relevance of the work presented in this study; the results of weakening the serratus anterior could inform muscle strength training for synergistic groups. Alternatively, the methods implemented to simulate weakness could provide information on the effects on joint forces. This is particularly relevant considering known associations between osteoarthritis (OA) and muscle weakness, discussed in detail in a review by Segal & Glass (2011). This review found multiple publications supporting the idea that weakness of the quadriceps was concurrent with OA incidence in the lower limb. A similar review by Loureiro et al. (2013) on hip OA found a number of studies reporting increased muscle weakness in the arthritic limb of patients relative to the contralateral limb. There appear to be no such publications on the association of upper limb arthroplasty and muscle weakness, however, the relationships seen in the lower body joints would likely apply to all joints considering the atrophy that may occur in older patients during a recovery period and extended disuse of their musculature. Rotator cuff tears reduce shoulder muscle strength; this study showed that weakness of the subscapularis,

supraspinatus and infraspinatus each caused significant increases in each other's peak muscle forces (other than weakness of the teres minor which had a negligible effect). For example, weakness of the subscapularis incited median increases of 67% and 49% in the infraspinatus and teres minor, and disabling supraspinatus incited 111% and 57% increases in infraspinatus and subscapularis. Rotator cuff tears become more prevalent with age (Tempelhof et al., 1999; Yamamoto et al., 2010), and therefore the effects of such a condition would be of interest during rehabilitation as muscle weakness symptoms would be enhanced in older populations.

There also appears to be a strong link between muscles classed as predominantly fast-twitch (containing a higher percentage of Type 2 muscle fibre) and increases in other muscle's activity when weakened. Those muscles which were highlighted as being fast-twitch, with the exception of coracobrachialis, all incited large increases in other muscle's activity. As discussed previously, the preferential degradation of this fibre type with age could indicate that weakness of these muscles is physiologically relevant in determining STS difficulties in the elderly. Figure 6.4 also indicated that weakness of these muscles significantly increased and/or decreased the peak glenohumeral and elbow joint contact forces. This would suggest reasons behind any increased discomfort in the upper limb during standing and sitting, and further work should investigate the relationship between a reduction in Type 2 muscle fibre and increased muscle weakness and joint pain.

The serratus anterior is highlighted as having the largest percentage of Type 2 muscle fibres in rat muscles – other data are not available - and as a point of reference it was shown that rodents have a higher proportions of Type 2 fibres in the vastus lateralis when compared to humans (Gregory et al., 2003). However, the similarities in ageing musculature with humans (Ballak et al., 2014) means this can serve as a good estimation of serratus anterior composition. Evidently, a reduction in the size and number of Type 2 fibres with age would have the most profound effect on the serratus anterior muscle. Considering the extreme effects of its weakness, this offers the possibility of one explanation into STS difficulties in the elderly. Without further validation, which would be possible if patients suffering from scapula winging could be recruited, this cannot be confirmed. However, there is strong evidence to suggest there is a link between serratus anterior weakness and problems in standing from a chair when

relying on arm rest support. Future work may look to set muscle force upper bounds and observe the effects on optimisation success, followed by working with scapula winging patients and tailoring rehabilitation to their specific needs.

6.4.1 Limitations

An un-constrained loadsharing optimisation led to the identification of pivotal muscles when muscle weakness was present in an extreme case. However, it also meant there were no set strength limits, leading to muscle forces values well above physiologically relevant values for the human body (particularly in the case of serratus anterior weakness). Future work should address the issues of subject-specific muscle force upper bounds, developing methods by which individual muscle lines of action, or functional groups (elbow/shoulder flexors and extensors for example), can be given bespoke upper bounds for more physiologically relevant optimisation constraints. One possible route would be through dynamometry tests, however, the number of movements required for a full-strength profile may prove impractical.

The profound effects of the serratus anterior on other muscles' peak levels of exerted force and joint reaction forces, along with the vulnerability to Type 2-specific atrophy with age, make this muscle an interesting link to difficulties in STS in the elderly. However, the only data available for serratus muscle fibre comes from a study on rats. Human data would be required to further support this hypothesis.

As discussed in the aims and objectives, this study was exploratory and did not focus on specific muscles. However, the outcomes of the work could lead to more informed muscle weakness studies where initial results, such as those presented here, can be used to identify muscles of importance. The work to be presented in Chapters 7 and 8 will also help to inform such work owing to the identification of pivotal singular, and groups of, muscles which are important to ADLs. These results would then be used to limit the work presented in this chapter, as the muscles disabled will be clinically relevant.

The aim of this study was to observe the effects of theoretical muscle weakness, it is therefore limited by the loadsharing optimisation of the MSK model used to

calculate the original muscle and joint force data. The effect of changing the cost function (sum of muscle stresses squared in the case of the UKNSM – see Chapter 4) on the ability to compensate for muscle weakness was not investigated, and altering the power or function itself may change the ability to produce successful solutions.

6.5 Concluding remarks

Kinematics measured from 27 healthy subjects performing a STS activity was analysed using the UKNSM musculoskeletal model with the effects of individual muscle weaknesses simulated. Of the 29 muscles omitted in this way, the serratus anterior proved the most debilitating, giving the largest percentage of the STS motion with no feasible loadsharing solution, and increasing the activity of other muscles and the upper body joint reaction forces by larger amounts than all other muscles. Previous studies have identified the debilitating effects of serratus anterior fatigue and weakness, however, this is the first study to demonstrate its importance during STS. The high prevalence of Type 2 muscle fibres in the serratus muscle suggests it could be vulnerable to age-related atrophy, thus suggesting a potential cause behind STS difficulties in the elderly. Due to its role in maintaining scapula stability during elevated activities, motions requiring large shoulder movements become painful and almost impossible when the serratus anterior is weakened. This may suggest possible routes for tailored rehabilitation techniques with future work looking for ways to identify local muscle weakness in individual patients. Other muscles with high proportions of fast-twitch muscle fibres were also shown to produce high levels of muscle force changes, further confirming the vulnerability of the elderly to the natural degradation of Type 2 fibres.

The clinical applications of this work could inform the effects of joint surgery on everyday activities, and the subsequent effects on muscle activity. This would require more detailed knowledge of subject-specific strength profiles for use in the MSK loadsharing optimisation. This would form a possible avenue for future research into muscle weakness.

Having investigated the effects of muscle weakness on young subjects, it falls to continue studying the effects of changing kinematics with age. This chapter looked at

comprehensively analysing upper body muscle weakness in young subjects, however different age groups may perform the STS motion with different strategies which may be more or less vulnerable to muscle weakness. The following chapter measures the STS activity for subjects of different ages, and how changes in strength with age may affect the biomechanics of activities of daily living.

CHAPTER 7

A BIOMECHANICAL ANALYSIS OF STANDING AND SITTING IN DIFFERENT AGE GROUPS

The previous chapter investigated the effects of theoretical muscle weakness on the sit-to-stand activity, but did not consider kinematic differences subject to weakening with age. This chapter presents the data obtained from the observation of different age groups standing and sitting from an instrumented chair. The setup, protocol and data analysis are described and results from MSK modelling presented.

7.1 Introduction

Current understanding of how different age groups perform activities of daily living is limited by the tools and data available. It is difficult to understand how varying movement strategies affect the whole body without a comprehensive analysis of all joints and muscles, and how strength deficiencies in older populations change the efficiency and comfort of everyday motions. Previous studies have mostly focused on one part of the body, or used estimations of joint and/or muscle power for biomechanical analysis. This study observed three different age groups performing a sit-to-stand activity, obtaining comprehensive kinematic and kinetic data through MSK modelling for the full body including external forces from the arm rests and seat of an instrumented chair, and further forces from the feet. The overall aim of this study was to quantify and understand the biomechanical changes during a sit-to-stand task between different age groups.

7.2 Methods

A power analysis was used to determine sample size (ClinCalc, 2017). Wheeler et al. (1985) investigated the muscle activation patterns of the triceps brachii in young and old women during standing and sitting, an activity which is representative of the work presented in this thesis. When standing from a standard chair, the average activity of the medial triceps brachii was found to be 28 ± 16.65 % (of maximum) and 39.8 ± 24.91 % (of maximum) in young and old respectively. Using this data to specify the means, standard deviations and differences between groups, an alpha value of 0.05 and power of 80%, the power analysis determined that a study with two independent groups and continuous data should include at least 30 subjects per group. As three different age groups are of interest in this study, a minimum of 90 subjects were required.

The protocol for capturing the sit-to-stand motion involved participants standing from and sitting in an instrumented chair as described in Duffell et al. (2013) (Figure 7.1). The seat height was kept at a set distance, 46 cm, from the ground throughout testing for better clinical relevance (Etnyre & Thomas, 2007; Millington et al., 1992;

Anglin & Wyss, 2000) than adjusting the height of the chair relative to knee or body height. Adjustments like this may allow comparison between individuals, but have poor translation to a clinical setting where seat heights are generally kept constant. The force sensors integrated into the chair used in this study consisted of one Kistler portable force plate (Kistler Holding AG, Winterthur, Switzerland) acting as the seat, two single point load cells (Vishay, model 1042-0050-G506R) to measure vertical hand forces and two tension/compression load cells (Omega, model LCM703-25) for the measurement of anterior-posterior (AP) hand forces. Previous published data on STS biomechanics suggested that the components of the exerted hand forces had different peak value onsets when comparing young and old (Alexander et al., 1991), however the values themselves were not presented. The instrumented chair allows the quantification of these values, and comparisons were possible between age groups. Medio-lateral forces at the hands were not measured. Two further Kistler portable force plates were placed in front of the seat to measure ground reaction forces for both feet. Movements were captured using a 10-camera optical motion capture system (Vicon Motion Systems, Oxford, UK), sampled at 100 Hz. Force plate and load cell data were sampled at 1000 Hz.

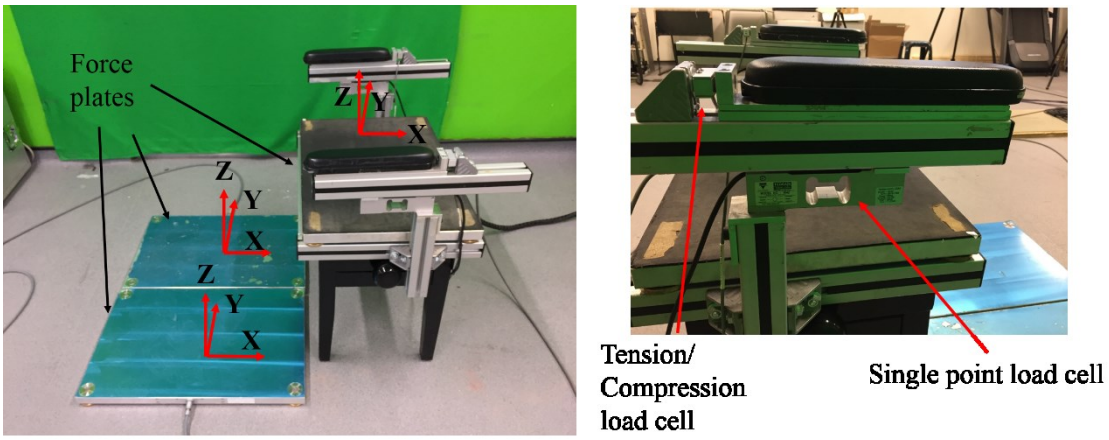


Figure 7.1 – Left: Instrumented chair with force plates. Coordinate systems for each force plate indicate positive directions for GRF components in each axis. Right: Instrumented arm rest. A positive force is defined as being in the superior (single point load cell) and anterior (tension/compression load cell) directions

The methods presented below were given ethical approval through the Joint Research Compliance Office and Imperial College Research Ethics Committee. The full application and approval letter can be found in Appendix 3.

95 males, divided into groups of young (YO, aged 18-35), middle-aged (MA, aged 40-60) and elderly (EL, aged 65 and over), participated in the study. Groups were defined based on age-related levels of strength as discussed in Chapter 2. The body is considered at peak physical strength after puberty, represented by the YO group, with no significant physiological changes until the 5th decade. The MA group therefore represent the beginning of functional decline. The EL group represents ages where declines in strength have already begun, and over 65 is the current boundary for an individual to be classed 'elderly'. With the proposed increases in retirement age, this definition may change in the near future.

YO and MA subjects were recruited from the general public and Imperial College London campus (South Kensington, London, UK) staff and students. EL subjects were recruited from the general public, with the majority being residents of the Royal Hospital Chelsea (London, UK). This facility is home to army veterans exclusively in post-retirement. All equipment was taken to the hospital location, where residents volunteered to take part.

After having explained the protocol and gained informed consent, participant heights and weights were measured and each was asked to provide information on hand dominance and any history of muscle/joint pain or surgery. They then completed the Edmonton Frail Scale, introduced and deemed acceptable in Chapter 3, which included a timed-up-and-go (TUG) test. Retro-reflective markers were placed on the arms, hands, torso, pelvis, legs and feet (Table 7.1), directly onto the skin using hypoallergenic double-sided tape and secured using Micropore surgical tape (3M Deutschland GmbH, Neuss, Germany).

Table 7.1 - Anatomical locations for reflective markers. * indicates markers removed after static and scapula measurements. Bracketed numbers correspond to positions shown in Figure 7.2

Segment	Markers	Anatomical location
Thorax	C7	7th cervical vertebra (19)
	IJ	Sternum jugular notch (14)
	MA	Sternum manubrium (16)
	PX	Sternum xiphoid process (17)
Pelvis	RASIS*	Right anterior superior iliac spine (1)
	LASIS*	Left anterior superior iliac spine (2)
	RPSIS*	Right posterior superior iliac spine (24)
	LPSIS*	Left posterior superior iliac spine (26)
	-	Cluster of three markers placed on the pelvis (25)
Clavicle	RAC* and LAC*	Right and left Acromioclavicular joints (12,18)
	RSC* and LSC*	Right and left sternoclavicular joints (13,15)
Scapula	-	Scapula trackers placed on both scapula spines (20)
Humerus	-	Cluster of three markers placed on the upper arm (21)
	LE*	Lateral epicondyle (22)
	ME*	Medial epicondyle (23)
Forearm	-	Cluster of three markers placed on the forearm (27)
	US*	Ulnar styloid (29)
	RS*	Radial styloid (28)
Hand	-	Cluster of three markers placed on the back of the hand (30)
	MCP2*	Second metacarpal (33)
	MCP5*	Fifth metacarpal (31)
Foot	FM2	Head of the second metatarsal (11)
	FCC	Calcaneus (32)
	FMT	Tuberosity of the fifth metatarsal (10)
	TF	Additional marker placed on the foot (9)
	FAM	Apex of the lateral malleolus (7)
	TAM	Apex of the medial malleolus (8)
Shank	-	Cluster of three markers placed on the shank (6)
Knee	FLE	Lateral femoral epicondyle (4)
	FME	Medial femoral epicondyle (5)
Thigh	-	Cluster of three markers placed on the thigh (3)

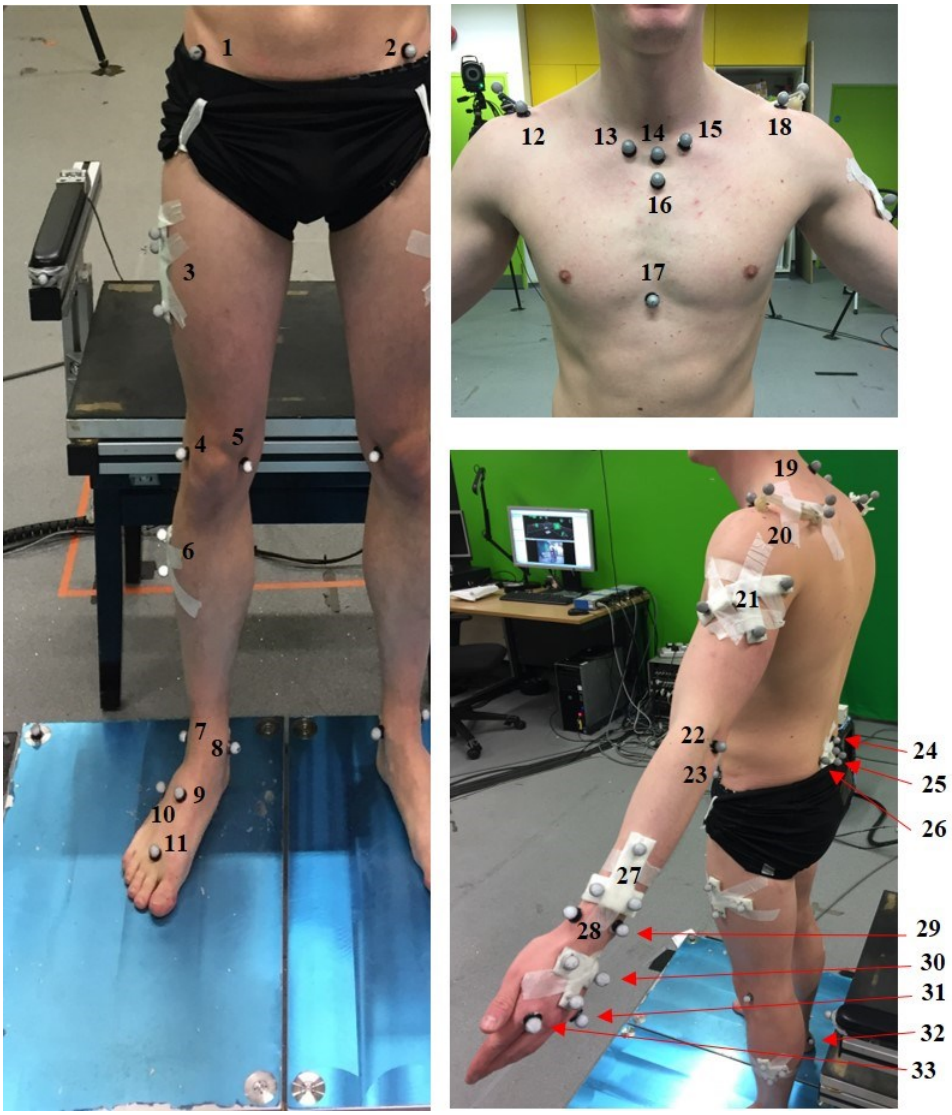


Figure 7.2 - All reflective markers positions. Details given in Table 7.1. Both sides of the body were tracked using the positions shown

Firstly, participants stood with one foot on each force plate whilst a static recording of the body was taken in the anatomical stance. Then participants sat in the instrumented chair with their ankles, knees, hips and back at approximately 90° (Figure 7.3). Once seated, calibrations of the scapula position and movements were performed on both the right and left hand sides (Figure 7.4) using the scapula tracker and locater described by Prinold et al. (2011). This device, shown in Figure 7.4, determined the locations of three scapula landmarks previously introduced in Chapter 2: Trigonum Spinae, Inferior Angle and Acromial Angle. Use of the tracker included a 30 second movement trial to determine the humerus rotational centre where participants were

asked to perform shoulder flexion and abduction, both with 90° of elbow flexion, and movements simulating the stirring of a large pot. They were then asked to stand and sit 6 times, if they were able, at a self-selected speed. For the first 3 trials, it was requested that they stand without using their arms or hands for support. No other instructions were given regarding hand position, so participants were free to move them during the trials, provided they did not use the armrests or knees for support. Participants were then asked to stand and sit a further 3 times but starting and finishing the motion with hands placed on the arm rests. No other instructions were given so each participant was permitted to stand as naturally as possible. Rests were given between repetitions when required.



Figure 7.3 – Experimental set up at the Royal Hospital Chelsea

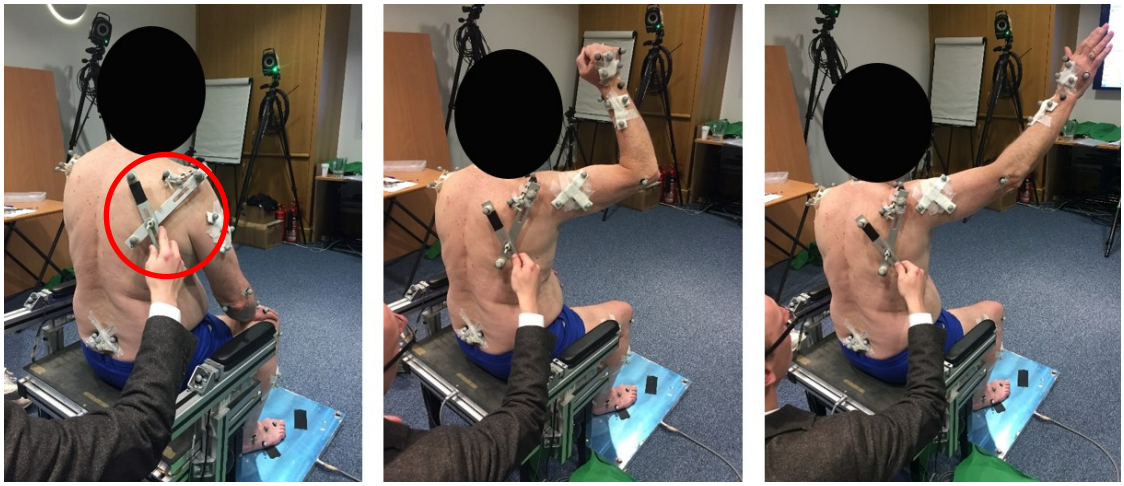


Figure 7.4 - The 3 static positions required for scapula calibration. Left - a resting pose with hands resting on the knees; centre – scaption (resting scapula position): shoulder at 90° abduction and elbow at 90° flexion with arm positioned approximately 30° to the coronal plane; right – shoulder at approximately 150° abduction. Scapula orientation was determined by the locator indicated by the red circle

Data was processed using Vicon-Nexus (Vicon Motion Systems, Oxford, UK) and Matlab (MathWorks, Natick, USA). Marker data was filtered using a 13th order low-pass Butterworth filter with a cut-off frequency of 4.7 Hz. This filter type and level of cut-off is in line with similar published work and literature recommendations (Gross et al., 1998; Hughes et al., 1996; Borstad & Ludewig, 2002; Cooper et al., 2002). Force plate and load cell data was also filtered using a low-pass Butterworth filter but with an order of 203 and cut-off frequency of 10 Hz (Winter, 2009). All data were input into the UKNSM and FreeBody (v. 2.1, Ding et al., 2016) for kinematic and kinetic analysis. Details of both models are given in Chapter 4.

Three external forces were measured and implemented into the MSK models: ground reaction force at the feet, reaction force of the seat (both for use in FreeBody) and measured hand loads at the arm rests (for use in the UKNSM). Use of individual force plates and loads cells for the feet and hands respectively gave data for the right and left-hand sides separately, however, only one force plate was used for the seat requiring the load through each side of the pelvis to be estimated. This was achieved by assuming that contact with the seat occurred through each ischial tuberosity of the pelvis

(as used by Wilson & Nicol, 2005) and that the measured reaction force was equally distributed through right and left contact points.

7.2.1 Subject-specific strength profiles

One prominent difference between inverse and forward dynamic MSK modelling is muscle contraction dynamics. In inverse dynamics, as the kinematics are measured, it only remains to quantify muscle forces. As described in Chapter 4, muscle forces are calculated using a mathematical optimisation algorithm which requires boundary constraints: the equations of motion with moment arms and intersegmental joint moments as the known variables, the glenoid boundary and muscle force upper boundaries (MFUB). These upper boundaries act as the solution space for the unknown variable muscle force, and are initially calculated using muscle PCSA and maximum muscle stress:

$$MFUB = PCSA * muscle\ stress \qquad \text{Equation. 7.1}$$

MSK models typically scale muscle morphology linearly from known anatomy, such as body segment parameters and muscle insertion/origin points. However, scaling MFUBs is not as straight forward and few studies have investigated the effects of such alterations. Instead of adjusting MFUB values to an individual, scaling muscle stress, after obtaining subject-specific PCSA values, has previously been explored (Bolsterlee et al., 2015). In Bolsterlee's study on improving MSK model predictions, the values for muscle stress were increased in steps of 1 Ncm^{-2} until a loadsharing solution was found; though no justification was given for the step size. Inter-subject variation in the final muscle stress values was between 13 and 18%. This method was presented as providing a more subject-specific strength profile, in comparison with the default model of set muscle stress and PCSA values for all subjects. However, measuring individual PCSAs is very time-consuming and Bolsterlee's work showed little difference between the muscle and joint force outputs from their generic and scaled MSK models. Studies by Knarr & Higginson (2015) and Wesseling et al. (2016) also worked to improve MSK modelling adjustments in maximum muscle force based on the torque generated during knee and hip dynamometry testing. This was shown to significantly improve joint contact predictions when compared to experimentally obtained data, but makes two

large assumptions without sufficient evidence of validity: 1 – muscles within the same functional group have similar levels of strength and can therefore be equally adjusted, and 2 – that subjects are producing maximal effort during isometric strength testing. In a similar study looking at tibiofemoral contact forces, Steele et al. (2012) applied arbitrary weightings (between 1 and 10) to the hamstrings, gastrocnemius and quadriceps' MFUBs. Their final loadsharing optimisation constraints were chosen once a combination of weightings resulted in differences between estimated and experimental knee contact forces of less than 20%. No justification was given for the range of 1 to 10, however, it would be logical to optimise the strength weightings based on differences between estimations and measured data. This assumes that the experimentally obtained data is reliable for comparison, and minimising differences does not necessarily mean accurate results if the comparative values remain dissimilar. Other studies that used integer multipliers to alter muscle boundary constraints include work by Persad (2016), who applied a multiplier of 5 to the MFUBs defined for use in the UKNSM , and that of Southgate et al. (2012), who doubled the upper bounds in accordance with the original conditions set within FreeBody. Prinold (2012) found that required multipliers (in order to get a solution) varied anywhere between 1 and 6 depending on the subject size and motion being analysed when using the UKNSM to model pull-up activities.

Developing subject-specific strength profiles may prove unnecessary depending on the sensitivity of MSK model outputs to changes in MFUBs. Southgate et al. (2012) found that doubling the PCSA values of 5 muscles in a lower-body MSK model, (which effectively increased the MFUB values), significantly increased the average number of successful solutions found for a jumping motion. They found that joint force values were not as sensitive as changes to kinematics when compared to muscles forces that were significantly different when the upper bounds were doubled. This shows that not all muscles in an MSK model need increased upper bounds, but knowing precisely which require adjustment is difficult to determine. The doubling presented by Southgate appears to be largely arbitrary and is presented as a sensitivity analysis. The applications of such testing in a clinical setting are unclear.

Many of these methods would be impractical for studies such as those presented in this thesis, as bespoke PCSA values will not be available and maximum isometric strength was not measured. However, there is still a need to represent the known

differences in strengths between different age groups (Chapter 2). Prinold (2012) found that when using the UKNSM, the MFUBs implemented did not always give an optimal solution to the muscle loadsharing optimisation. They eventually unbounded the optimisation. This is not physiologically accurate, but does reveal which muscles are of greatest mechanical advantage in creating the intersegmental moments. If this thesis is to quantify MSKR, a more logical method of scaling will be required. This can be addressed using the review in Chapter 3, which discussed existing research on elderly ADL performance and concluded that older subjects employed strategies that resulted in joint torque requirements close to their maximum strength capacity.

The MFUBs, as optimisation boundaries, define the largest amount of tension muscles can produce. This in-turn defines the current strength profile used to distribute the load from the calculated joint moments across the available muscles. The boundaries as calculated may not always provide a solution where constraints are not broken and the cost function is minimised (Prinold, 2012; Southgate et al., 2012). A failure in the optimisation would indicate that the model could not find a solution using the available muscle strengths, and therefore indicates that the loadsharing optimisation requires a larger solution space. Physiologically speaking, this would mean that the motion being analysed requires greater muscle forces than the model allows. It therefore stands to reason that, assuming these constraints are restricting the muscles in producing ample force to give the required joint torques, gradually increasing these boundaries and finding the minimum strength profile giving a solution would indicate the body working at its maximum potential.

The increments by which multipliers of MFUBs are increased has not previously been discussed in detail by the literature. The simple act of requiring scalar multipliers indicates inaccuracies in the initial calculation of Equation 7.1. This subsequently suggests errors in the values used for PCSA and maximum muscle stress, both of which have been shown to vary between muscles (Buchanan, 1995), age and gender (O'Brien et al., 2010), anatomical studies (Maganaris et al., 2013; Narici et al., 1992) and those reportedly used in MSK models (Prinold, 2012). The reported variation in either of these variables could indicate a minimum step size between multipliers, which would account for any potential errors in predicting the MFUBs.

Prinold (2012) reviewed the use of maximum muscle stress in MSK models in the literature and compiled a list of implemented values between 33 and 120 N/cm^2. The UKNSM uses a value of 100 N/cm^2 in its prediction of MFUBs, therefore comparing the values reported by Prinold with this gives a variation of approximately 20%.

This error in maximum muscle stress could be used in the calculation of physiologically-based, subject-specific strength profiles. This is possible in combination with the points discussed so far: 1 – the elderly perform activities of daily living at, or very close to, their maximum capacity in terms of strength and 2 – MSK models will not solve mathematical optimisation without sufficient solution space, which is limited by maximum strength per muscle. If the MFUBs (by way of a scalar multiplier) were continually increased until a successful solution was found, then the minimum multiplier giving a solution may indicate the absolute minimum strength required to perform the motion. The potential error in calculating these upper bounds could be in the region of 20%, as per the variations found previously through Prinold's review of maximum muscle stress. Therefore, in this study simulations were initially run using no multiplier and systematically increased in steps of 0.2 until a solution was found. The multiplier was applied globally, so all muscles simulated within the MSK models will have upper boundaries multiplied by the same number. The assumption here is that previous optimisation failures were caused by insufficient muscle force allowance in some muscles, and that any successful solution will be found by giving the optimisation the minimal required solution space. This therefore simulates the body working at maximum capacity, at least for the muscles requiring larger MFUBs. It would be unusual for every muscle in the body to be working at maximum capacity, as not every movement requires all muscles to be maximally active. This means that, using the concept applied above, if one or a group of muscles is working at maximum capacity then it may be the case that other muscles can synergise the movement if required. Implementing a global increase means these compensatory muscles have an additional increase in MFUB, despite the fact that they were not previously 100% active.

The review of performing activities of daily living close to maximum capacity was only found for an elderly population, therefore the approach of incrementally increasing MFUB multipliers was only implemented in the EL group as finding the

minimum strength for an optimisation solution is only physiologically relevant for this age.

If it is assumed that the elderly maximum strength levels are equal to the minimum effort required, then the maximum force profile for the MA and YO groups can be estimated according to the optimum EL scalar multiplier. Subject-specific strengths were not measured in this study, however, the review in Chapter 2 on changing muscle strength with age can be used. It was found that on average the elderly had 25 and 37% reductions in estimated strength levels compared to that of young subjects for the upper and lower body muscles respectively. Applying these differences to the calculated EL multipliers for both MSK models (UKNSM and FreeBody) increased the maximum muscle force for the MA and YO groups, simulating higher strength levels. As the differences in strength are mean population values, and subject-specific strengths were not measured, then the adjusted multipliers for the younger groups were applied to all subjects. A lack of sufficient data separating young and middle-aged subjects in terms of strength meant both groups were assumed to have similar strengths.

7.2.2 Phase segmentation

For the current study, a two-phase approach to analyse the sit to stand motion was adopted. This took into account the phases used in previous studies (Chapter 3) and combines the use of force plate data and angular velocities. Phase 1 starts with trunk movement initiation, ending at a point just as the subject leaves the seat and the force plate acting as the seat gives a value of 0 N. Having left the seat, Phase 2 consists of the movements up until hip flexion angular velocity first reaches 0°/sec, having initiated just after lift-off. These two phases make up a whole 'Stand' motion. The phases of sit have not been previously determined, and therefore the same events as stand will be used. Phase 3 begins at initiation of hip flexion but from a standing position, and ends as the subject contacts the seat at which point Phase 4 begins (when the seat force first increases from 0 N). The instrumented chair was backless, and so there is no measure of resting on the back of the chair, therefore Phase 4 ends when the trunk reaches its most posterior point.

7.2.3 Data analysis and statistics

All data was tested for normality using the Shapiro-Wilk's test ($p>0.05$). Two-way mixed ANOVAs were used to determine any significant interactions between age and use of arm rests for durations, ground reaction forces (GRFs), all joint reactions forces (JRFs) and peak muscle forces. In the absence of any interactions, the ANOVA tests also highlight the main effects of age or use of arm rests. One-way ANOVAs will be used to examine differences between age groups for arm rests forces.

Concerning the upper limb: when grouping all 29 peak muscle forces for each subject, the data was found to be non-normally distributed. Therefore, a Mann-Whitney U test was used to compare the median values between age groups.

7.3 Results

As this thesis is looking to explore the whole-body biomechanical changes in completing a sit-to-stand motion for different age groups, the following results sections will explore both upper and lower limb joint and muscle data with and without the use of arm rests. Three of the EL group were unable to stand up without the use of armrests and have been excluded from the analysis as no comparisons can be made between using and not using armrests. A further 6 were only able to perform STS twice without arm rests. Only data from the right-hand side of the body was analysed in this thesis (83% of participants stated that they were right-hand dominant). Average values were taken across all trials for each subject, separating the trials with and without the use of arm rests. The following sections will first look at the time taken for each age group to complete the STS task. Data from the upper and lower bodies will then be presented separately, first looking at the progression from external forces (on the arm rests or through the feet) to joint reaction forces through the shoulder, hip and knee. The associated muscle forces for both sections of the body will then be presented.

For motion durations, total stand (both Phases 1 and 2) and sit (both Phases 3 and 3) will be presented along with individual phase data. In subsequent sections and

tables, the terms 'Stand' and 'Sit' will denote the combination of Phases 1 and 2 and Phases 3 and 4, respectively.

All data are presented as maximum peak values. For lower body muscle and joint force data, these peaks typically occurred just after leaving the seat. However, in some cases (standing with the use of arm rests for example), a distinct peak at this point in the trials was not clear. In these cases, an alternative reference point was used – quadriceps activity - which consistently presented with clear peaks just after leaving the seat.

7.3.1 Participant information

Participant data are summarised in Table 7.2. The EL group were significantly shorter than both YO and MA groups ($p<0.0005$). There were no differences in height between YO and MA groups. The MA group had a significantly greater body mass than the YO group ($p<0.05$) only.

Table 7.2 - Participant data

Group (n)	Age, years (± SD)	Height, m (± SD)	Body mass, kg (± SD)	Frailty Score (median score, min, max)
YO (30)	28 (5)	1.80 (0.07)	78.6 (12.4)	0 (0,4)
MA (30)	52 (6)	1.79 (0.07)	89 (12.7)	0 (0,3)
EL (35)	78 (7)	1.72 (0.06)	83.4 (11.6)	2 (0,7)

7.3.2 Motion durations

Only Phase 1 and the total stand (Phases 1 and 2) presented with significant differences between age groups: EL were significantly faster when using arm rests compared to YO ($p<0.05$) during Phase 1 and took longer than MA when not using arm rests ($p<0.05$) for Phase 1 and total stand. All groups showed a significant increase in the duration of the total sit task when using arm rests ($p<0.0005$, Table 7.3).

Table 7.3 - Duration of whole motion and individual STS phases for all age groups. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05). [m] (p<0.05) or [M] (p<0.005) indicates significance differences between using and not using arm rests within an age group (applicable to all groups when placed after phase name)

Mean duration in seconds (± SD)

	With arm rests			Without arm rests		
	YO	MA	EL	YO	MA	EL
Total stand	2.08 (0.39)	1.96 (0.26)	2.10 (0.41)	2.08 (0.30)	1.94 (0.23)	2.26 (0.59)[b]
Total sit [M]	2.27 (0.38)	2.43 (0.49)	2.42 (0.50)	1.98 (0.34)	2.13 (0.32)	2.12 (0.45)
Phase 1	0.69 (0.18)	0.64 (0.13)	0.59 (0.15)[a]	0.69 (0.11)	0.64 (0.11)	0.75 (0.23)[b,m]
Phase 2	1.39 (0.28)	1.32 (0.23)	1.38 (0.39)	1.39 (0.27)	1.30 (0.19)	1.48 (0.41)
Phase 3 [M]	1.47 (0.31)	1.57 (0.36)	1.61 (0.41)	1.21 (0.25)	1.29 (0.24)	1.32 (0.37)
Phase 4	0.73 (0.17)	0.79 (0.21)	0.69 (0.21)	0.69 (0.17)	0.75 (0.15)	0.73 (0.18)

7.3.3 Muscle force upper boundary multipliers

The mean multiplier required to give successful solutions (EL only) to the loadsharing optimisation was 2.5 (± 1.2) for the UKNSM and 0.43 (± 0.1) for FreeBody. Applying the differences in strength between age groups, as reported in Chapter 2, resulted in group multipliers of 3.4 and 0.68 for the UKNSM and FreeBody respectively for both YO and MA groups. However, a feasible solution (defined as the loadsharing optimisation being able to produce muscle forces within the constraints set by the MSK model) was not always found. In these cases, the originally determined multipliers of 3.4 and 0.68 were gradually increased in steps of 0.2 (as with the EL group's original algorithm) until a solution was found. This gave FreeBody multipliers in the range of 0.68 to 1.2 for both YO and MA and UKNSM multipliers in the range of 3.4 to 4.4 (with two extreme cases of 5.2 in YO, and 4.6 in MA) for both YO and MA.

7.3.4 Upper body

The UKNSM load-sharing optimisation was unable to find loadsharing solutions for 3 YO, 5 MA and 2 EL group members. Therefore, the group sizes presented here are 27, 25 and 30 respectively for the upper limb.

7.3.4.1 Arm rest forces

Table 7.4 shows the peak hand forces, relative to body weight, for both stand and sit. As each arm rest contained two load cells, the data are presented as peak resultant and components: vertical and anterior-posterior (AP). No significant differences were found between any age group for both stand and sit portions of STS. Onset of peak hand force (Table 7.5) showed no significant differences between age groups.

Table 7.4 - Mean peak force exerted on arm rests. Vertical is positive in the superior direction and AP is positive in the anterior direction. There were no significant differences.

	Mean peak force (% body weight ± SD)								
	Resultant force			Vertical component			AP component		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	18.9 (6.1)	18.7 (7.9)	22 (5.8)	18.3 (6.6)	18 (8)	21.1 (5.3)	-4.4 (2.7)	-5.4 (3.1)	-5.5 (1.9)
Sit	18.4 (5.1)	17.3 (7.6)	20.3 (4.9)	17 (5.5)	16.4 (7.4)	18.7 (4.4)	-5.9 (3)	-5.3 (2.4)	-6.8 (2)

Table 7.5 - Onset of peak resultant hand forces. There were no significant differences.

	Time of peak resultant force (% trial length from start ± SD)		
	YO	MA	EL
Stand	38 (6.2)	36.4 (6.3)	34.5 (6.5)
Sit	67.8 (5.5)	68.3 (6.8)	67.2 (7.5)

7.3.4.2 Shoulder Joint Reaction Force

Shoulder JRF (glenohumeral JRF) is presented in three parts: resultant force, resultant shear force (both AP and superior-inferior components of the JRF) and medial-lateral (ML) force. The UKNSM defines the ML force as being positive in the lateral direction, therefore a positive value indicates a reaction to the compression of the humeral head into the glenoid. Each force is given in the scapula anatomical frame and is constrained to be within the rim of the glenoid cavity. Table 7.6 shows the data for peak JRF when arm rests were used and Table 7.7 presents the same data for when arm rests were not used. When using arm rests, the EL group was found to be significantly larger than the YO group in all JRF components during stand and sit barring the shear force during sit. This also extended to significant differences to MA in sit. When not using arm rests, significant differences were only found between MA and EL during stand for ML force, and between YO and EL during sit for resultant and ML force. No significant interaction was found between the use of hands and age on shoulder JRF during stand, but a significant interaction was found for sit ($p<0.05$). Concerning the resultant shear force, there was a significant interaction between age and use of arm rests in both stand and sit ($p<0.05$). A significant interaction was seen between age and use of hands on ML force during sit only ($p<0.05$).

Table 7.6 – Peak shoulder joint reaction force during stand and sit when using the arm rests. [ab] indicates significant differences to the YO/MA groups respectively ($p<0.05$)

	Mean peak JRF (% body weight ± SD)								
	Resultant force			Resultant shear force			ML force		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	133.8 (61.2)	170.9 (114.1)	206.4 (96.5)[a]	73.4 (31.8)	85.1 (45.3)	104.9 (48.1)[a]	111.2 (54.7)	147.5 (108.1)	177.9 (90.1)[a]
Sit	179.9 (86)	185 (114.9)	267.9 (120.4)[ab]	98.8 (39.2)	85.8 (45.8)	126.6 (61.1)[b]	146.9 (68.8)	162.1 (110.6)	234.4 (114.3)[ab]

Table 7.7 - Peak shoulder joint reaction force during stand and sit without using arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05). All values, for all groups, were significantly smaller than when using arm rests (p<0.0005)

Mean peak JRF (% body weight ± SD)

	Resultant force			Resultant shear force			ML force		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	23.8 (11.1)	22.5 (6.4)	27.5 (8.7)	6.4 (3.8)	7.3 (3.7)	6.9 (3.2)	23 (10.5)	21 (5.1)	26.6 (8.6)[b]
Sit	21.3 (5.8)	23.6 (9.9)	27.5 (8.3)[a]	5.6 (2.1)	6.8 (3.3)	7.2 (3.8)	20.7 (5.5)	22.7 (9.6)	26.5 (8)[a]

7.3.4.3 Upper limb muscle forces

For comparison between subjects, and considering the MFUB scalar multipliers implemented prior to the loadsharing optimisation, muscle force will be presented relative to body weight (BW) as opposed to muscle activation (force relative to MFUB), which will be introduced in Chapter 8. Only trials where the arm rests were used will be presented, as the previous section showed the JRF during the 'without arm rests' trials to be negligible compared with the 'with arm rests' scenarios. Therefore, any associated muscle forces will not be of interest for the current study.

Peak muscles forces for the upper body muscles are presented in Table 7.8. Of the 29 muscles featured in the UKNSM, 10 were significantly greater in EL compared to YO and/or MA, with the anterior portion of the deltoid presenting as significantly higher in YO, during sit, and 5 muscles were significantly greater in EL compared to YO during stand. The peak force from the anterior portion of the deltoid and medial head of the triceps significantly decreased from YO to MA.

Table 7.8 - Peak muscle forces for the upper body whilst using arm rests. [ab] indicates significant EL differences to the YO/MA groups respectively ($p<0.05$, uppercase $p<0.0005$). Only muscles with at least one significant difference between groups have been presented. Full results can be found in Appendix 4. Full muscle names can be found in the glossary of abbreviations and Chapter 6

	Mean peak muscle force (% body weight ± SD)					
	Stand			Sit		
Muscle	YO	MA	EL	YO	MA	EL
LS	10 (9.2)	9.2 (9.1)	13 (12.8)	16.6 (14)	12.7(13.1)	24.3 (21.5)[b]
Rmin	4.4 (3.6)	3 (3.5)	4.5 (4.5)	7 (5.6)	3.7 (4.3)	8.3 (9.6)[b]
Pmin	4.1 (2.9)	5.1 (4.9)	7.4 (6.5)[a]	7 (5.6)	6.7 (7.1)	9.8 (7.7)
DeltA	26.5(15.8)	15.9(20.8)	19.1 (16.9)	35.7(26.4)	17.7(22.2)[a]	17.1 (16.2)[a]
SS	24.6(19.7)	35.8(31.3)	44.8 (36)[a]	19 (14.1)	36.3(31.5)	51.4 (40.3)[a]
Tmaj	33 (22.6)	49.9(36.1)	63.2(35.9)[a]	58.3(31.9)	61.4 (43)	90.1(45.4)[ab]
BIC_long	30.7(25.4)	28.8(26.9)	46.7 (27)[b]	37.5(24.6)	28.5(24.9)	58.7(27.1)[aB]
TRI_long	18.5(17.4)	31.9(25.4)	37.8(26.6)[a]	27.7(25.7)	35.5(26.2)	52.6 (35.5)[a]
TRI_med	46.3(35.8)	41.6(30.9)	51 (29)	67.6(41.7)	43.7(34.7)[a]	69.4 (28.2)[b]
TRI_lat	26.1(21.4)	24 (19)	29.9 (17.9)	40.7 (29)	25.5(20.8)	43.6 (21.4)[b]
Anc	8.8 (8)	8.7 (7.7)	13.6 (12.5)	13.7 (9.7)	10.7 (9.7)	22.9(14.4)[ab]
BRD	1.3 (1.1)	1.2 (1.3)	0.9 (0.8)	1.3 (0.8)	1.7 (1.8)	0.9 (0.7)[b]

7.3.5 Lower body

Lower limb data will be presented in the same format as the upper limb. FreeBody was unable to produce a feasible solution for one subject within both the YO and MA groups. Therefore, the total number of subjects for each group were 29, 29, and 32 for YO, MA and EL respectively.

7.3.5.1 Ground reaction forces

Force plates under each foot allowed the acquisition of ground reaction forces (GRFs). The scenario of using arm rests did not always present with a clear defined maximum peak, as shown in Figure 7.5, and so the alternative reference point (peak knee extensor activity) was used to define the point of having just left the chair. This was also an issue for the sitting trials, where a similar form of reference point was used.

Tables 7.9 and 7.10 contain data for the peak resultant forces, as well as vertical and AP components. As movements were predominantly in the sagittal plane, medial-lateral components of the GRF were negligible and are not included in this analysis.

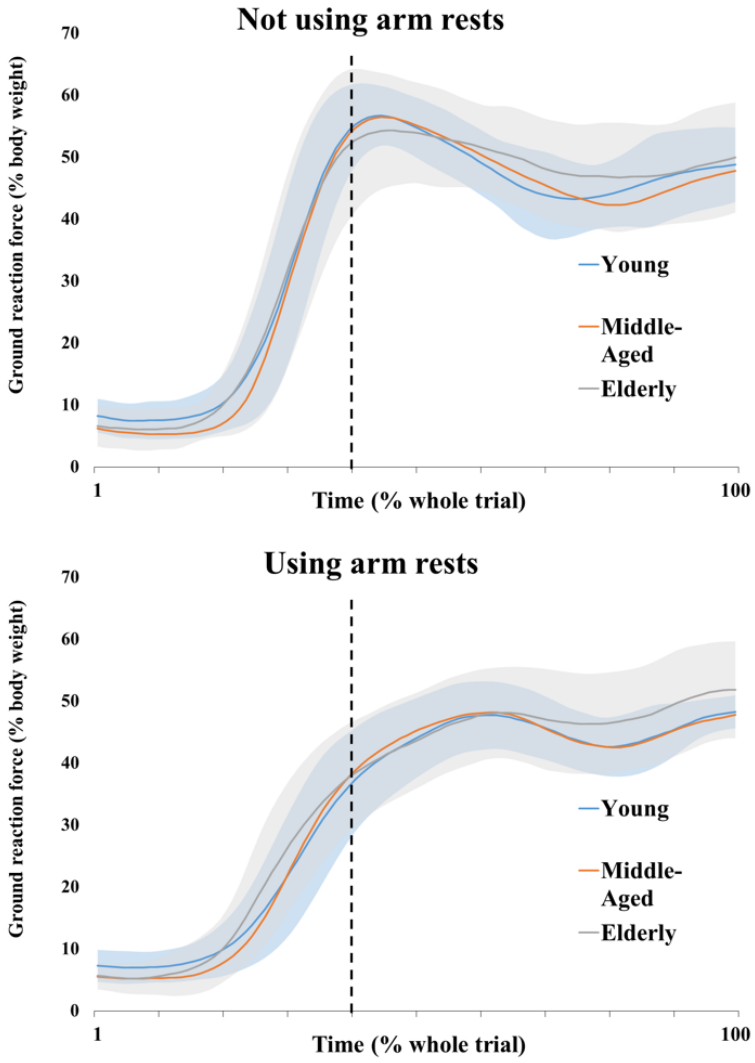


Figure 7.5 - Top: ground reaction force profiles for all three age groups when standing and not using arm rests (coloured areas show standard deviations for YO and EL, MA not included). Bottom: ground reaction force profiles when standing and using arm rests

Table 7.9 - Peak ground reaction forces when armrests were used. [ab] indicates significant EL differences to the YO/MA groups respectively (p<0.05)

Mean peak force (% body weight ± SD)

	Resultant force			Vertical component			AP component		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	32.5 (6.2)	30.7 (8.1)	27.9 (8.2)	32.3 (6.2)	30.5 (8)	27.7 (8.1)	1.1 (0.7)	1.4 (0.7)	1.9 (1.1)[a]
Sit	35.9 (7.1)	30.3 (8.4)[a]	30.5 (8.6)[a]	35.8 (7)	30.2 (8.3)[a]	30.4 (8.6)[a]	0.8 (0.7)	1.1 (0.9)	1.8 (1.4)

Table 7.10 - Peak ground reaction forces when armrests were not used. All components for all groups were significantly larger than when arm rests were used (p<0.0005). [ab] indicates significant EL differences to the YO/MA groups respectively (p<0.05)

Mean peak force (% body weight ± SD)

	Resultant force			Vertical component			AP component		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	57 (8.6)	58.3 (8.6)	54.8 (9.2)	56.8 (8.6)	58 (8.5)	54.6 (9.1)	4.1 (1.4)	5 (1.8)	5.3 (2.1)[a]
Sit	55.9 (7.8)	54 (6.8)[a]	50.5 (9.5)[a]	55.7 (7.8)	53.8 (6.7)[a]	50.3 (9.5)[a]	4.9 (1.6)	4.7 (1.7)	5.2 (2)

No significant interaction was found between use of hands and age for both stand and sit. Significant differences were predominantly in sit, where the resultant forces and vertical components were significantly smaller in the MA and EL compared to YO. EL were also found to have significantly larger peak AP components during stand against YO. All groups showed a significant decrease in peak GRF when using the armrests during stand and sit ($p<0.0005$ for all groups) compared to not using arm rests. The same significant findings also apply to the vertical (no significant interaction between use of hands and age, and significant decrease in peak vertical GRF when using arm rests, $p<0.0005$) and AP (no significant interaction between use of hands and age, and significant decrease in peak AP GRF when using arm rests, $p<0.0005$) components.

7.3.5.2 Onset of peak ground reaction force

The time during the motion at which the peak GRF occurs may indicate changes in important events between age groups. Peak times are measured as a percentage of the whole motion and are presented in Tables 7.11 and 7.12.

Table 7.11 - Onset of peak GRF when using the arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05)

	Time of peak resultant force (% trial length from start ± SD)		
	YO	MA	EL
Stand	49.7 (15)	45.8 (16.4)	37 (12.9)[ab]
Sit	50.5 (8.5)	59.1 (14.5)	63 (12.7)[a]

Table 7.12 - Onset of peak GRF when not using the arm rests. YO was significantly different for both stand (p<0.05) and sit (p<0.0005) when compared to using arm rests

	Time of peak resultant force (% trial length from start ± SD)		
	YO	MA	EL
Stand	40 (5.8)	43.3 (6.5)	40 (8.3)
Sit	60.4 (6.1)	57.8 (8.3)	61.3 (9.3)

No significant interaction was found between age and use of armrests on peak GRF onset during stand or sit, but significant differences were found between using and not using arm rests in the young for both stand and sit. During stand, the peaks were earlier in the motion when not using hands and delayed for sit. When using the arm rests, the peak GRF for EL occurred significantly earlier than both the MA and YO groups in stand, and YO during sit. This delay was approximately 12% of the motion (compared to YO) during stand and 13% during sit.

7.3.5.3 Knee contact forces

JRF at the knee will be presented in four parts: total tibial-femoral (TF) JRF, medial and lateral JRF components of TF JRF and patello-femoral (PF) JRF. TF and its components are presented in Table 7.13 (with arm rests) and Table 7.14 (without arm rests).

Table 7.13 - Peak knee contact force when arm rests were used. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase p<0.0005).

Mean peak JRF (% body weight ± SD)

	Total TF force			Medial compartment force			Lateral compartment force		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	239.2 (68.7)	235.5 (48.4)	209.9 (41.6)	108.6 (24.7)	101.5 (19.6)	84.6 (24.6)[ab]	138.5 (54.9)	139.1 (38.3)	132.8 (33)
Sit	236.4 (57.1)	227.3 (40.5)	215.2 (25.1)[a]	110 (22.1)	100.8 (19.1)	87.5 (17.9)[Ab]	134.4 (45.8)	130.7 (33.5)	132.4 (26)

Table 7.14 - Peak total knee contact force when arm rests were not used. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase p<0.0005). All values for all groups were significantly greater than when using arm rests (p<0.0005)

Mean peak JRF (% body weight ± SD)

	Total TF force			Medial compartment force			Lateral compartment force		
	YO	MA	EL	YO	MA	EL	YO	MA	EL
Stand	317.1 (75.7)	313.4 (74.1)	280.5 (58)	141.7 (36.2)	121.7 (27)	103.6 (32.1)[Ab]	184.3 (60.8)	195.1 (54.2)	185.3 (45.3)
Sit	297.3 (78)	276 (43)	255.3 (40.9)[a]	137.4 (35.5)	114.2 (17)	101.7 (22.1)[A]	167.8 (56.8)	165.2 (35.6)	161.5 (35.8)

No significant interaction between use of arm rests and age was found on total JRF or lateral compartment force during stand or sit. A significant interaction was found for medial compartment force for both stand and sit (p<0.05). In all scenarios where a significant difference was noted, the EL group saw decreases in JRF compared to both YO and MA groups. When comparing the individual compartments of the TF joint, the medial side saw a significant decrease in JRF however the lateral compartment saw no change with age. This applied to both with and without the use of arm rests.

7.3.5.4 Patello-femoral contact forces

PF forces are presented in Table 7.15 (with arm rests) and Table 7.16 (without arm rests). A significant interaction was found between use of arm rests and age for both

stand and sit ($p<0.0005$). Both the MA and EL groups had consistently lower values for peak PF joint force in all scenarios when compared to YO.

Table 7.15 - Peak patello-femoral contact forces when using arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase: p<0.0005)

	Mean peak JRF (% body weight ± SD)		
	YO	MA	EL
Stand	520 (268)	376 (143)[a]	274.4 (132.9)[Ab]
Sit	452 (215)	315.7 (122.3)[a]	242.2 (91.6)[A]

Table 7.16 - Peak patello-femoral contact forces when not using arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase: p<0.0005)

	Mean peak JRF (% body weight ± SD)		
	YO	MA	EL
Stand	850.6 (357.9)	584.7 (209.8)[a]	426.4 (166.3)[Ab]
Sit	774.5 (328.6)	524.3 (193.1)[A]	377.2 (153.6)[Ab]

7.3.5.5 Hip contact forces

JRF at the hip joint is presented in Table 7.17 and Table 7.18. A significant interaction between use of arm rests and age was found for sit only ($p<0.05$). A significant increase in hip JRF was seen when not using arm rests ($p<0.0005$) in all groups for stand and sit compared to using the arm rests. Significant decreases in hip JRF was seen between YO and EL groups.

Table 7.17 - Peak hip contact forces when using arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05)

	Mean peak JRF (% body weight ± SD)		
	YO	MA	EL
Stand	260.5 (56)	239 (54.2)	214.9 (53.5)[a]
Sit	257.4 (41.2)	248.4 (57.5)	234.3 (43.2)

Table 7.18 - Peak hip contact forces when not using arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05)

	Mean peak JRF (% body weight ± SD)		
	YO	MA	EL
Stand	416.4 (107.2)	367 (65.8)	349.2 (79.2)[a]
Sit	374.5 (96.5)	334.4 (64)	301.4 (63.1)[a]

7.3.5.6 Lower limb muscle forces

As with the upper body muscles, the lower body peak muscle forces are presented below. However, this will include trials where the arm rests were not in use. Data are presented as muscle forces relative to body weight, in Tables 7.19 and 7.20. Full muscle force data can be found in Appendix 4.

The EL group were found to have a large number of significant decreases in peak muscle force over multiple muscles, with larger differences when no arm support was used. During stand, comparing YO to EL, the older group presented with 6 and 8 muscles with significant increases in peak force whilst using and not using arm rests respectively. This compares to 15 and 18 muscles with significant decreases. Similarly in sit, EL show 2 (with arm rests) and 4 (without arm rests) muscles with increased force against 16 and 20 muscles with decreased force. Considering all age groups, the decreases in peak muscle force were largest in the vasti and rectus femoris muscles in both stand and sit, with hamstring muscles (namely semimembranosus and the long head of biceps femoris) and ankle plantarflexors (gastrocnemius and soleus) producing the largest increases with age.

Table 7.19 – Peak muscle forces for the lower body during stand. [ab] indicate significant differences to the YO/MA groups respectively (p<0.05, upper case p<0.0005). m (p<0.05) or M (p<0.0005) indicate significant difference between use of arm rests; if placed after muscle name significant differences were found for all groups. Only muscles with at least one significant difference between any age group have been presented

	Mean peak muscle force (% body weight, SD)												
	Stand			Stand – no arm rests				Stand			Stand – no arm rests		
Muscle	YO	MA	EL	YO	MA	EL	Muscle	YO	MA	EL	YO	MA	EL
Add.magn. dist.[M]	38.7 (8.8)	31.5(7.6)[a]	26.3 (9)[A]	53.9(10.3)	46 (11)[a]	42.4 (12)[A]	Psoas major	8.7 (4.9)	15.1 (6.6)[a]	15.4 (7.9)[a]	10 (5.7)	17.6 (13.4)[a]	14.8 (9.1)[a]
Add.magn. mid.[M]	20.7 (4.7)	17.3(4.3)[a]	14.4(4.6)[A]	29.3 (6.5)	26.1 (5.8)[a]	23.9 (6)[A]	Rectus fem.[M]	84.6 (51.4)	52.1 (21.4)[a]	38.9 (21.2)[A]	140.6 (71)	86.5 (34.1)[A]	62.3 (26.5)[A]
Bic. fem. CL	22.3 (12)	27.3 (16.3)	31.6 (16.5)	25.1 (13.3)	41.4 (30.7)[a,m]	52.1 (24.4)[A,M]	Sartorius (prox.)[M]	4.3 (3.9)	6.3 (4.6)	5.6 (3.4)	6.7 (5.2)	11.3 (7.3)[a]	11.2 (4.8)[a]
Ext. dig. long.[m]	4.4 (5.2)	1.9 (2.5)[a]	1.5 (1.3)[A]	6.7 (7.2)	2.6 (3.6)[a]	1.9 (1.6)[A]	Semimembr	11.4 (5.2)	13.5 (5.9)	15.3 (6.2)[a]	12.7 (5.7)	18.6 (11.6)[a,m]	23.8 (10.3)[A,M]
Ext. hal. long.	5.3 (6.2)	2.7 (3.2)	1.8 (1.4)[a]	10 (10.3)[m]	3.5 (4.4)[a,m]	2.1 (2)[A]	Semitend.[M]	31.7 (23.4)	20.7 (10.5)[a]	16.1 (5.8)[A]	62.2 (48.7)	34.5 (21.5)[a]	28.4 (10.2)[A]
Gastrocn. (med.)	12.9 (11.9)	13.6 (8.9)	23.6 (16)[ab]	14 (15.7)	11.5 (6.4)	24.8 (26.5)[ab]	Soleus (med.)[M]	21.1 (19)	36.2 (27.9)	58.8 (43.1)[A]	32.6 (29.7)	50.3 (37.1)	65.6 (42.5)[A]
Gracilis[M]	5.3 (3.6)	3.6 (1.5)[A]	2.7 (1.2)[a]	13 (8.5)	7.1 (3)[A]	5.7 (2.6)[A]	Tensor fasc. l.	6.5 (4.2)	12.1 (7.9)[a]	11.9 (7.4)	7.2 (5.9)	17.1 (25.7)[a]	11.5 (9.3)
Iliacus (lat.)	1.7 (1)	3 (1.3)[a]	3 (1.6)	2.4 (2.2)	3.5 (2.6)[a]	3 (1.8)	Tibialis ant.	22.4 (21.3)	25.4 (19.4)	16.8 (12.7)	42.2 (36.5)[M]	40.2 (31.3)[m]	17.5 (13.3)[ab]
Iliacus (mid.)	4.7 (2.7)	8.2 (3.6)[a]	8.4 (4.3)[a]	5.8 (3.9)	9.6 (7.3)[a]	8.1 (5)[a]	Vastus interm.[M]	75.5 (47.9)	45.6 (20.1)[a]	32.1 (19.3)[A]	143.5 (79.8)	74.8 (34.2)[A]	51.2 (31.6)[A]
Iliacus (med.)	2.1 (1.2)	3.7 (1.6)[a]	3.7 (1.9)[a]	2.9 (2.4)	4.3 (3.3)[a]	3.7 (2.3)[a]	Vastus lat. (inf.)[M]	8.5 (5.7)	5.8 (2.2)[a]	4.2 (2.1)[A]	17 (12.3)	9.2 (3.6)[a]	7.2 (4.6)[A]
Peroneus brev.	17 (15.4)	9.1 (10.2)[a]	5.6 (6.1)[A]	29.4 (26.5)[m]	10.2 (11.8)[A]	7.6 (8.4)[A,m]	Vastus lat. (sup.)[M]	229.7 (119.7)	136.8 (58.2)[A]	96.7 (53.8)[A]	329.8 (158.8)	199.6 (87)[A]	133.1 (66.4)[A]
Peroneus long.	8.7 (5.9)	9 (6.6)	7.5 (6)	19.1 (23.9)[m]	13.3 (14.4)	7.4 (7.4)[a]	Vastus med. (inf.)[M]	10.9 (8)	9.4 (3.3)	7 (3)[a]	16.3 (11.8)	14.9 (5)	11.7 (5.3)[a]
Peroneus tert.	5.3 (6.9)	3 (4.3)	1.7 (1.9)[a]	9.9 (10.5)[m]	3.4 (5.5)[a]	2.4 (3.1)[A]	Vastus med.(mid.)[M]	50.5 (29.9)	36.7 (13.1)[a]	26.5 (12.6)[A]	88.8 (45.2)	59.1 (20.5)[a]	44.7 (18.8)[A]
Popliteus	5.4 (7.6)	2.8 (2.3)	3.3 (3.5)	10.9 (15)[m]	3.7 (3.9)[a,m]	4.8 (7)[a]	Vastus med. (sup.)[M]	42.5 (25.7)	27.2 (10.1)[a]	19.9 (9.6)[A]	81.9 (44.3)	44.9 (17.2)[A]	35.2 (17.4)[A]

Table 7.20 - Peak muscle forces for the lower body during sit. [ab] indicate significant differences to the YO/MA groups respectively (p<0.05, upper case p<0.0005). m (p<0.05) or M (p<0.0005) indicate significant difference between use of arm rests; if placed after muscle name significant differences were found for all groups. Only muscles with at least one significant difference between any age group have been presented

	Mean peak muscle force (% body weight, SD)												
	Sit			Sit– no arm rests				Sit			Sit– no arm rests		
Muscle	YO	MA	EL	YO	MA	EL	Muscle	YO	MA	EL	YO	MA	EL
Add. magn. (dist.)[M]	38.7 (8.1)	30.3 (6.9)[A]	26.7 (6.6)[A]	51.3 (11)	40 (8.2)[A]	35.9 (10.2)[A]	Rectus fem.[M]	72.2 (41.4)	47.5 (19.1)[a]	33.3 (11.7)[A]	132.1 (71.1)	77.8 (32.1)[A]	56.5 (26)[A]
Add. magn. (mid.)[M]	20.8 (4.5)	16.8 (3.9)[A]	14.7 (3.4)[A]	27.8 (6.3)	22.6 (4.2)[A]	20 (5)[A]	Sartorius (prox.)[M]	4.6 (3.9)	7.2 (5.2)[a]	6.6 (2.9)	6.4 (5.2)	10.5 (6.2)[a]	9.8 (4.8)
Add. magn.(prox.)[M]	2.5 (0.7)	2.3 (0.8)	1.9 (0.5)[ab]	3.9 (1.2)	3.6 (1)	3 (1)[ab]	Semimembr	16.1 (6.8)	17.4 (6)	18.2 (4.6)	15.6 (5.7)	19.3 (7.8)[m]	21.5 (7.4)[a,m]
Bic. fem. CL	29.5 (16.7)	35.2 (17.6)	37.9 (13.2)	28.5 (14.2)	41.1 (23)[m]	46.3 (21.5)[a,m]	Semitend.[M]	25.9 (16.9)	18.7 (7.2)[a]	16.1 (3.9)[a]	55.7 (48.9)	31.7 (20.6)[a]	23.1 (8)[A]
Ext. dig. long.	5.3 (5.9)	2 (2.3)[A]	1.7 (1.6)[A]	6.8 (7.4)	1.9 (2.8)[A]	1.9 (1.6)[A]	Soleus (med.)[m]	19.7 (18.9)	33.5 (26.3)	58.1 (41.3)[Ab]	28.1 (28)	41.1 (35.1)	56.9 (38.5)[Ab]
Ext. hal. long.	5.6 (5.3)	2.6 (2.9)[a]	1.9 (2.1)[A]	9.6 (12)[m]	3.1 (3.8)[a]	1.8 (1.9)[A]	Tensor fasc. L[m]	6.6 (4.5)	10.2 (5.7)[a]	9.2 (5.5)	5.9 (4.7)	9.1 (6.5)[a]	7.9 (6.6)
Flex. dig. long.	0.7 (0.8)	0.8 (0.9)	0.8 (0.5)	2 (5)	1 (1.5)	1 (0.7)[m]	Tibialis ant.	25.3 (17.9)	23.4 (17.6)	15.7 (12.5)	38.9 (31)[m]	34.6 (24.9)[m]	13.9 (12.7)[Ab]
Gastrocn. (med.)	11 (13.1)	11.9 (9.3)	19.2 (14.1)[ab]	10.9 (13)	10.9 (9.1)	21.3 (21.4)[ab]	Vastus interm.[M]	63.8 (35)	38.3 (15.8)[A]	27 (12.8)[A]	125.2 (73)	66.7 (32.3)[A]	43.5 (27.4)[A]
Gemellus (inf.)[M]	2 (0.7)	2.2 (0.9)	1.9 (0.7)	3.4 (1.8)	3 (0.8)	2.5 (0.9)[a]	Vastus lat. (inf.)[M]	7 (3.4)	4.9 (1.6)[a]	3.7 (1.4)[A]	13.9 (10.7)	8 (3)[a]	5.8 (2.8)[A]
Gemellus (sup.)	1.9 (0.5)	2 (0.7)	1.9 (0.6)	2.9 (1.4)[m]	2.5 (0.7)[M]	2.3 (0.7)[a,M]	Vastus lat. (sup.)[M]	202.1 (103.4)	114 (50.2)[A]	83.8 (42.3)[A]	321.5 (156.9)	182.8 (84.1)[A]	122.7 (65)[A]
Gracilis[M]	4.2 (2.6)	3.2 (1.3)	2.3 (0.8)[A]	10.3 (7.5)	6.1 (2.7)[A]	4.4 (1.8)[a]	Vastus med. (inf.)[M]	9.2 (4.2)	8.5 (2.6)	6.7 (2.2)[a]	15.1 (11.1)	13 (3.8)	10 (3.9)[a]
Peroneus brev.	16.2 (13.3)	8.2 (9.7)[a]	5.6 (8.2)[a]	25.1 (24.9)[m]	8.8 (12.3)[a]	6.1 (6.9)[A]	Vastus med.(mid.)[M]	43.7 (21)	32.3 (10.5)[a]	24.6 (8.7)[A]	78.3 (42.6)	52.6 (17.9)[a]	38 (15.7)[A]
Peroneus tert.	6.4 (6.9)	3 (4.1)[a]	1.8 (2.7)[a]	9.4 (11.6)	3 (4.8)[a]	2.1 (2.8)[a]	Vastus med. (sup.)[M]	36 (17.4)	24 (8.4)[a]	18.1 (6.3)[A]	69.8 (39.2)	40.2 (15.6)[A]	28.9 (13)[A]

7.4 Discussion

7.4.1 Durations

Durations for the YO and EL groups are in good agreement with previously published data for total stand, Phase 1 and Phase 3 for both using and not-using arm rests (Alexander et al., 1991). Phase 1 data from Alexander and colleagues found the young to stand in 0.59 (± 0.15) and 0.63 (± 0.15) seconds with and without using arm rests respectively, and a group of elderly subjects (classed as being able to stand without arm rests) were found to rise in 0.65 (± 0.17) and 0.84 (± 0.41) seconds. These values are comparable with the current study (YO – 0.69 & 0.69 seconds, EL – 0.59 & 0.75 seconds), however the values from Alexander's works for Phase 2 and total stand were marginally higher. This may be a result of their methods in segmenting the motions, which differ from the current thesis work, and are more ambiguous in their descriptions. The end of Phase 2 is denoted as reaching a 'fully upright configuration' whereas this thesis used a more quantifiable variable (0 °/sec hip angular velocity) which may have been achieved prior to being in an upright position.

There is also agreement with studies showing that stand-to-sit takes longer than chair rise (Mourey et al., 1998). EL were found to complete Phase 1 significantly faster than YO, albeit with an absolute difference of only 0.1 seconds. There were no significant differences between YO and EL groups when not using arm rests, with some differences seen between the latter and MA. However, all ages performed the whole activity and individual phases within very similar time frames. All events can therefore be considered to have taken place at the same time within all groups. This may reflect the fitness of the EL group, who were predominantly army veterans who spent a large amount of time in physiotherapy or exercising. The EL group was rated on average as 'Not Frail' according to the Edmonton Frail Scale. Groups rated as more frail may have taken significantly longer to perform STS.

7.4.2 Upper body external forces and joint reaction forces

Few studies have assessed as wide a range of ages in tandem, and presented both resultant and component external forces for the hand. The MA vertical component results (18 ± 8 % BW) are in good agreement with the control subjects of Duffell et al. (2013, 15.5 ± 7.6 % BW) and the resultant forces (18.7 ± 7.9 % BW) with mixed gender results of subjects aged 51 to 64 years old (Anglin & Wyss, 2000, 24 % BW, average 16 %). Anglin & Wyss showed that resultant hand forces were lower in sit than stand by 3% BW (significance unconfirmed), though neither designation of side, nor whether the data are for one or two sides, was stated. The young and elderly resultant data (18.9 ± 6.1% BW and 22 ± 5.8 BW respectively) are both larger than data published by Alexander et al. (1991), who showed that both young and old men exhibited around 25% body weight for a summation of both right and left hand sides during chair rise. The population size used by Alexander and colleagues was a fraction of the sample presented here and differences may simply lie in the participants used in each study.

There are several significant differences between all three groups in terms of the kinetics of the STS movement for the shoulder. The JRFs were consistently higher in the EL group when compared with both YO and MA despite no differences in external force applied on the arm rest. Onset of peak hand force was also similar between age groups. Again, the fitness of the EL group may suggest they are capable of rising without heavy reliance on the arm rests. A lack of significant differences in AP components suggest all groups were predominantly pushing vertically down at peak showing no evidence of the EL using the chair to accelerate themselves forward as one might expect given the need to reach a stable position quickly (Alexander et al., 1991). The increase in upper body JRF in the EL group suggests that there might be increased discomfort when performing the STS. The reasons behind these differences will be further explored in Chapter 8.

In all groups, using the arm rests increased shoulder JRF dramatically. Only one other study could be found reporting predicted shoulder contact forces during standing and sitting, and only when using arm rests. Anglin et al. (2000) found that for a group of male and female adults (aged 51 to 64 years), the contact forces were found to be 1.8 (times BW, SD ± 1) for stand and 1.3 (times BW, SD ± 1) for sit. This is comparable

with the YO and MA groups (1.3 times BW and 1.7 times BW respectively for stand and 1.8 times BW for both YO and MA for sit), with some extremes of Anglin's data reaching 4 times BW. This is restricted to peak resultant forces, where the work in this thesis has also shown significant differences between age groups for shear and ML forces at the shoulder, and presents data for arms-free movement. The data when not using arm rests were all significantly smaller than whilst using arm rests, and fall well below the values reported for other ADLs such as driving, lifting weights or planar movements (Westerhoff et al., 2009; Anglin et al., 2000; Bergmann et al., 2007; Klemt et al., 2018).

7.4.3 Lower body external forces and joint reaction forces

The peak ground reaction force data for the lower body showed no significant differences between any age group during stand for resultant and vertical component forces. However, EL had increased AP components during stand and decreased peak resultant and vertical values during sit, both relative to YO. Onsets of the peak values did show significant differences when using arm rests between EL and both YO/MA groups. When taking into account these onsets, and the lack of significant difference in motion duration, it is clear that the EL group began transferring weight to the lower body earlier than the younger participants during stand. For sit, the EL are delayed in terms of peak GRF. These profiles are less defined than for stand, but may suggest the EL wait until they have sufficiently lowered their CoM before shifting their weight to the seat. This may increase stability or confidence in elderly movement.

The peak values for the right leg (54.8 – 57 % BW) are in good agreement with previously published data on peak GRFs during STS when arm rests are not used (in the range of 56 to 59 % BW - Millington et al., 1992; Gross et al., 1998; Etnyre & Thomas, 2007) but lower than the literature when hands were placed on arm rests (27.9 to 32.5 % BW in the current study compared to 49.9 % BW in Etnyre & Thomas, 2007). The onset times of peak GRF are within 10 % (approximately 0.2 seconds) of other studies where the participants stood without arm rests (Etnyre & Thomas, 2007; Millington et al., 1992).

Few studies exist in the literature which predict or quantify loads in the knee during STS. In all cases the literature data is based on older subjects, and therefore, considering the elderly subjects of this study, the lower end of each range given for the current study is used as comparison. The current study found TF force to be between 2 and 2.4 times BW when arm rests were used and 2.8 to 3.2 times BW when not used, which is in agreement with published data from instrumented implants (2.46 times BW when not using arm rests, Kutzner et al., 2010) and other MSK models (~ 3 times BW, Trepczynski et al., 2012). The work by Trepczynski failed to state whether arm rests were used or not. Comparison with the current study, and work by Kutzner and colleagues, show the research done by Trepczynski may not have used arm rests. Trepczynski et al. (2012) also used a MSK model to predict PF joint forces, with STS producing peak JRFs of approximately 3.1 BW. This is smaller than the values of 8.5 to 4.3 times BW presented in this thesis. Trepczynski's participants (only 2 were used) both had a history of total knee replacement (which will only have measured TF, not PF, force), and the kinematics and kinetics cannot be assumed comparable to that of a healthy group of subjects (Alnahdi et al., 2011; Hatfield et al., 2011; Levinger et al., 2013).

No literature could be found for comparisons against instrumented implants for the PF joint, however, Reilly & Martens (1972) used a 2D planar model to estimate the forces in this joint during extreme degrees of knee bending. For a maximum knee flexion angle of approximately 130-140°, it was predicted that loads could reach up to 7.6 times BW. This is comparable with the values of the YO group (8.5 times BW) in the current study, owing to the large degree of initial knee flexion during STS, but does not take into account the accelerations and large peaks in GRF associated with leaving the chair without the support of arm rests. The results presented here can be considered the first published for the PF joint within multiple age groups with and without the use of armrests.

The hip joint contact forces (3.5 to 4.2 times BW in stand, 2 to 3.7 times BW in sit) are larger compared to data from instrumented implants during stand and sit (1.9 and 1.6 times BW in stand and sit respectively, Bergmann et al., 2001) but good agreement, particularly in sit, with another modelling study (an average 2.6 times BW for both stand and sit, Varady et al., 2015); in both cases there was no arm rest support.

These publications, as with studies on the knee, feature older subjects (and so comparisons are made to the minimum for each range presented above for the current study), whereas the current study presents data for multiple age groups and during different arm support scenarios. The study by Bergmann and colleagues only included 4 patients, and the reported inter-subject variation makes it difficult to make fair comparisons. Differences in strategy may prevent high loads in patients that have gone through hip (and knee) arthroplasty, essentially protecting joints and reducing the measured *in vivo* loads presented in the instrumented implant study. Previous studies have found that newly operated patients load their non-operated side (after unilateral hip arthroplasty) more than the operated/effected limb (Kubonova et al., 2016; Martínez-Ramírez et al., 2014; Talis et al., 2008).

An MSK simulation study by Varady featured 11 patients aged around 40 years old, with data that shows the same trend (as presented in this thesis) for hip JRF being lower in sit compared to stand. There is also some agreement in MA and EL peak hip contact force (3.7 and 3.5 times BW respectively during stand) when Varadys' maximum values are considered (~ 3 times BW during stand).

When comparing the upper and lower body JRFs whilst using arm rests, it is clear the elderly are relying more on the upper body to provide most of the required lifting force which is consequently increasing the amount of JRF in the shoulder. Significant increases in the shoulder force, coupled with significant decreases in the PF force and quadriceps activity, show that using arm rest support appears to reduce the force on the knees and quadriceps but in the long-term may be causing further pain and discomfort in the upper body due to fatigue or overuse. Increased muscle forces show that the elderly are exerting higher amounts of effort to achieve this, but the difference in strategy seems logical owing to a naturally greater decline in lower limb strength with age (Chapter 2) when compared to the upper limb. It is necessary to further explore the changing pattern of muscle forces which may help in explaining the increase/decrease in JRFs.

7.4.4 Whole body muscle forces

The data for the upper limb show the EL group produced a larger amount of peak force during both stand and sit in several muscle groups despite no significant differences in external forces applied at the hands. It would be logical to assume the increased muscles forces are acting to stabilise the higher JRFs seen at the shoulder. Increased muscle forces around a joint usually indicate stabilisation, something which is known to increase with age due to a growing desire to stiffen joints for more stable movements (Hortobágyi et al., 2009; Hortobágyi & Devita, 2000; Klein et al., 2001).

When analysing individual muscle forces, the biceps and triceps in particular stand out as contributing more force per body weight in the EL group than the younger participants, in both stand and sit. Previous studies have shown an increase in co-contraction when compared to the young for these muscles during simple planar movements (Bautmans et al., 2011), and a review by Hortobágyi & Devita (2006) showed this to be consistent across of a range of muscle groups and movements. Co-contraction, also termed co-activation, increases joint stiffness and stability. In terms of STS, this would suggest increased stability of the elbow whilst lifting the upper body from the chair. The supraspinatus and teres major also presented with elevated peak activity in the EL group. Shoulder extension is important in projecting the body out of the chair whilst using arm rests support, therefore an increase in teres major force would suggest an increase in the amount of push the elderly apply to the arm rests. Shoulder stability is highly important in many upper limb activities, and increased supraspinatus activity may again indicate an increase in coactivation as the shoulder adducts in stand. Kinematics of the shoulder will be further explored in Chapter 8, and may help to understand the differences seen between young and old.

Combining the increases in upper limb muscle activity with the significant decreases in lower limb muscle force shows the significant effect of using arm rest support on the whole body. Considering the natural reduction of lower limb strength in the elderly, it would seem logical that the ageing body has a preference to load stronger muscles to achieve daily activities. However, the full extent to which this affects the upper body joints was not fully known until now.

For the lower body, different findings to the upper body emerge. When considering the trials without arm rests: during the stand motion 8 muscles produced a significantly greater peak force for EL compared to YO and 18 produced significant reductions in peak force. It is much the same with sit, where 20 EL muscles produced significantly less peak muscle force than YO and 4 produced significantly higher peak forces. Focusing on the stand trials, and performing a Mann-Whitney U test on the groupings of all YO and EL peak muscle forces which were found to be significantly different (including all increases and decreases from Tables 7.19 and 7.20), revealed the median values not to be significantly different (YO – 18.1% BW, EL – 16.2% BW, $p = 0.249$). It was also found that YO had 3 peak muscle force outliers, as determined by boxplot, of 140.6, 143.5 and 329.8% BW (rectus femoris, vastus intermedius and the superior portion of vastus lateralis – all quadriceps muscles). Knowing the similarity in medians and inter-quartile ranges (IQR - YO 40.8%, EL 33.4%) would suggest that the EL are exerting an overall similar amount of peak muscle force in order to achieve the same motion. Knowing the large difference in maximum muscle strength between the young and old (Chapter 2) means the elderly are exerting a relatively larger amount of effort, and performing closer to the maximum capacity. The large knee extensor outliers in the YO group explain the large differences in patello-femoral JRF, however it also shows that the younger groups rely more heavily on knee extension to complete chair rise. It would appear that in the EL group, the reduction in vasti activity is balanced by an increase in hip extensors and ankle plantarflexors. This is essentially prioritising protection of the knee, but may be sacrificing the integrity of the hip and ankle.

The decrease in lower body JRF discussed in the previous section was said to suggest a protection of the joint, and previously discussed changes in peak muscle force supports this. For the PF joint, a significant decrease in quadriceps (rectus femoris and vasti muscles) peak muscle force in the EL group clearly explains why the JRF is reduced (owing to the location of their insertion points on the patella). The decrease in hip JRF for both stand and sit whilst not using arm rests may be in part due to the significant decrease in semitendinosus (stand: YO 62.2 & EL 28.4% BW, sit: YO 55.7 & EL 23.1% BW) activity, although both the semimembranosus (stand: YO 12.7 & EL 23.8% BW, sit: YO 15.6 & EL 21.5% BW) and long head of biceps femoris (stand: YO 25.1 & EL 52.1% BW, sit: YO 28.5 & EL 46.3% BW) presented with significant increases. The effect in comparison to the PF joint forces are proportionally far less, and

may be a result of changing torque requirements. This, along with potential changes in joint kinematics and positioning of the body during the STS motion, could explain the inconsistencies between JRF reduction and muscle force increases with age.

When considering all JRF and muscle forces together, a distinct lack of consistent significant differences between MA and both YO and EL suggests a bridge between the youngest and eldest group which the MA participants appear to be occupy. In many cases, such as lower body peak muscle forces and PF joint forces, significant differences exist between YO and MA, in addition to the differences seen between YO and EL and between MA and EL. A lack of significance will always suggest similarities between groups, however, in these cases it appears as though the MA group represent the boundary between healthy and declining function. The age of the MA group suggest, as with many studies found in the literature (Chapter 2), that this decline begins around the $5^{th}/6^{th}$ decade. The lack of significance between YO and MA would suggest the older of these two groups are still able to perform the STS motion without compromising joint integrity. Significant differences in muscle activity, in lower limbs and to a certain extent the upper limbs, suggest the movements require a higher level of muscular effort. Almost every lower limb significant difference found between YO and EL was accompanied by an additional significant difference between YO and MA. In the vasti muscles particularly, this showed steps between YO and EL, and in reality, could be showing signs of changing muscle recruitment strategies to maintain a healthy functional ability prior to age-related decline. This intermediate stage of the MA group would suggest that it may be possible to identify early changes that can inform early rehabilitation and training interventions. These are discussed in Chapter 8.

7.4.5 Limitations

Despite numerous publications on the elderly performing ADLs at maximum capacity, the current group may exceed others of the same age in terms of muscle strength. The cohort used in this study were predominantly military veterans who regularly exercised and may be considered above average in terms of health and fitness. This group may also have a history of joint overuse due to the extreme activities associated with their military backgrounds. Statistically significant differences were still seen between age groups, suggesting a shift in strategy to accommodate weakening

muscles in order to successfully complete STS. Others of the same age, who do not stay as active, may not be able to complete the task at the same speed as those featured in the current study, and therefore differences between groups may be amplified. Some of the participants were unable to complete the activity, and had to be excluded from the dataset. Future work may look to quantify these attempts, and identify the point at which they failed. Comparison with the dataset in this thesis may locate important points of STS where difficulties are amplified.

Findings from Chapter 2 were used to quantify differences in strength between young and old subjects based on published data on dynamometry tests. There was limited information for those featured in the middle-aged group, however, it was assumed that these subjects would have a strength level equivalent to that of the youngest group. Discussions from Chapter 2 would suggest this to be a logical assumption, as age-related declines in muscle size and strength are most prolific around the sixth-decade. However, some studies suggested decline could begin as early as 30/40 years old. To address this uncertainty, future research should quantify subject-specific strength levels as an estimation of the group's strength. This may also help to inform more bespoke muscle force upper boundaries, although this would be difficult owing to the number of modelled muscles featured in both the UKNSM and FreeBody.

As the instrumented chair contained load cells for the measurement of hand loads in only two directions, ML components could not be measured. As the participants were restricted to movements predominantly in the sagittal plane, these ML forces are expected to be low however no published data is available to confirm this. Future developments of the instrumented chair should consider the addition of multi-axial load cells to quantify hand forces in all directions.

It was not possible to produce muscle force-loadsharing solutions for small numbers of subjects when using both the UKNSM and FreeBody. When using MSK modelling, it can sometimes be the case that the equations of motion produce intersegmental forces and moments which the optimisation algorithm is unable to share amongst the available muscle lines of action. In these cases, the scaled anthropometrics for muscle moment arms may not be representative of the subjects in question. This could be a result of using different anatomical datasets (giving poorly scaled muscle

origin and insertion points), the variation of which has been shown to alter modelling outputs (Ding et al., 2016). For this thesis, unsuccessful solutions were not used. However, future investigations may look at the possibility of using different datasets or even altering the cost function (see discussion in Chapter 6) in light of solving for these anomalies. Sensitivity analyses could be then used to determine any differences between the results presented in this thesis and any new optimisation parameters.

7.5 Concluding remarks

Three groups of men of different ages performed a sit-to-stand motion whilst external forces and movements were observed using an instrumented chair, multiple force plates and an optical motion capture system. Musculoskeletal modelling, with customised muscle force upper boundaries, was used to quantify muscle and joint reaction forces for the upper and lower parts of the body.

When comparing the data for all three age groups, it was found that significant differences existed for a variety of different biomechanical aspects of the STS motion. Joint reaction forces for the shoulder were generally elevated in the elderly group, despite no such differences in external forces at the hands in magnitude or onset of peak values. Greater peak forces in the elderly upper body muscle groups partly explained the increase in joint force, and suggest the older groups were working harder to achieve the same motion as their younger counterparts.

Studies have previously shown that the elderly lose strength in their legs prior to weakening of the upper body. This is reflected in the findings of this thesis. The peak muscle and joint forces of the YO group were significantly higher than the EL group for the knee, hip and multiple lower body muscle groups. This was particularly evident in the knee extensors, where corresponding increases in the hip extensors and ankle plantarflexors in the elderly suggest a change in muscle recruitment strategy. These findings would suggest a protection of the patello-femoral joint in particular, with the possibility that the elderly routinely alter movement strategy to reduce the force on this part of the body. When considering the MA group, there were inconsistencies in the significant differences to both YO and EL. The JRFs tended not to be significantly

different other than in the patello-femoral and quadriceps forces however multiple muscles showed a significant difference between YO and MA, particularly in the lower limbs. This would suggest a point at which the body is able to perform ADLs, albeit at a higher cost of muscle activity, without significant discomfort to joints despite elevated JRF values relative to EL. JRFs may start to increase/decrease as a result of this, eventually leading to the extreme changes in recruitment strategy seen in the elderly.

Having analysed the biomechanical changes between age groups for joint and muscle forces, the next chapter will begin to explore the changes in muscle activation (muscle force relative to upper bounds as opposed to body weight) and the kinematic changes. This will help in explaining why the differences found in this chapter exist between age groups and how all the changes are a consequence of a reduced musculoskeletal reserve.

CHAPTER 8

QUANTIFYING THE EFFECTS OF MSKR ON SIT-TO-STAND PERFORMANCE

This chapter further discusses the findings from Chapter 7, exploring the effects of a changing MSKR on the young and elderly groups. Both kinematics and muscle activations are presented.

8.1 Introduction

Chapter 7 presented a comprehensive exploration of the biomechanics of sit-to-stand in various age groups, showing key points as to changes in both muscle and joint reaction forces for multiple parts of the body. For the upper body of the elderly group, this highlighted significant increases in peak shoulder and muscle forces compared to the young group. The data for the elderly lower body showed decreases in both hip and knee joint forces (compared to the young), accompanied by a dramatic change in muscle recruitment strategy.

Changes in joint reaction forces, owing to the general lack of significant differences found between groups for external forces at the hands and feet (Sections 7.3.4.1 and 7.3.5.1 respectively), may be due to either kinematic changes or the new distribution of muscle forces found in the elderly. It was found that alterations in muscle recruitment in the elderly lower limb muscles resulted in different functional groups exhibiting higher amounts of activity (Tables 7.19 and 7.20), namely the ankle plantarflexors and hip extensors.

The aim of this chapter was therefore to assess the kinematics of both the upper and lower body, as differences in joint reaction forces may be derived from alterations in movement strategy or speed. Required joint torques at the ankle, knee and hip were analysed, in order to better understand the consequences of a redistribution of muscle forces away from the quadriceps in the elderly.

Movements during STS occur predominantly in the sagittal plane, therefore only movements in this plane will be presented in the following sections. Independent t-tests were used to assess significant differences between age groups. This chapter also looked at the changing levels of muscle activation (muscle force relative to maximum force boundary) for each age group. This quantifies the changes in MSKR, and how the alterations in movement strategy are ultimately affecting the vulnerability of the elderly. If the chosen motion is resulting in higher levels of overall activation, then it would be logical to assume that continued weakening of ageing muscle would eventually prove detrimental to functional ability.

The scenarios of using and not using arm rests were explored. However, they are separated into upper and lower body sections: the upper body will be used to discuss scenarios where arm rests were used, whereas not using arm rests will be analysed using the data from the lower body. This is due to the profound effects of each scenario on the respective areas of the body.

8.2 Whole body Kinematics

The aim of this section is to explore the kinematics of the shoulder, trunk and lower limbs to better understand any changes in movement between groups. As the majority of statistically significant differences were found between the young and elderly groups in Chapter 7, only these will be presented in full.

The discussion on postural stability in Chapter 3 showed that positioning the body's CoM is important in the elderly for achieving a successful chair rise. This change in CoM trajectory relative to the hip (Tables 7.17 and 7.18) and knee (Tables 7.13 to 7.16) may well explain the decreases in lower body JRF from the young to the elderly, as well as changes in muscle force (Tables 7.19 and 7.20). Therefore, the position of joints relative to ground reaction force centre of pressure will be assessed to find differences in postural stability between age groups.

The results of Section 7.3.4 showed that, despite similarities in peak external hand forces applied at the arm rests, the elderly had significantly higher resultant joint reaction forces at the shoulder. Two possible reasons behind this may be inter-group variations in shoulder width (increasing the moment arm between arm rests and the glenohumeral joint) or in shoulder kinematics. Table 8.1 shows the anatomical data (distance between right and left scapula acromion), where the elderly group were found to have significantly narrower shoulders. As each subject was asked to sit centrally in the chair, and the width of the arm rests was consistent, then this means that the moment arm of the external force to the shoulder joint in the elderly group was larger.

Table 8.1 - Shoulder width as defined by the distance between right and left acromion. The elderly were found to be significantly smaller ($p<0.0005$) than the young.

Age group	Shoulder width in cm (± SD)
Young	34.8 (2.3)
Elderly	31.6 (3.2)

The absolute differences in Table 8.1 are small, and two alternative theories may explain the differences in GH forces: movement of the clavicle or positioning of the arm prior to leaving the chair. Figure 8.1 presents the clavicle elevation during sit-to-stand whilst using arm rests. The peak values of the elderly group were significantly larger than the young group. This shows that the shoulder complex will have been more elevated prior to leaving the chair, which may have led to the significantly larger force seen acting through the shoulder in the elderly.

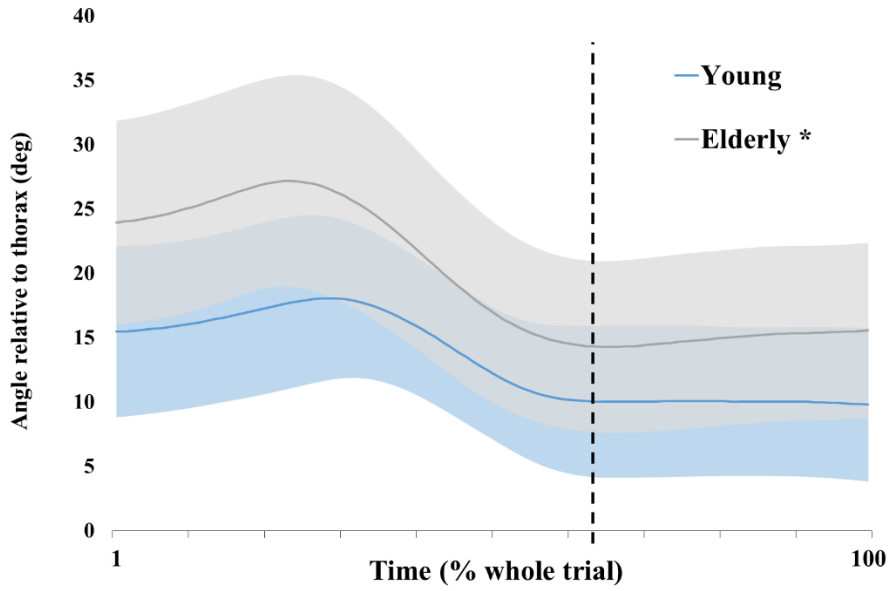


Figure 8.1 – Clavicle elevation whilst performing STS with the use of arm rests. Time at which the hands leave the arm rests is indicated by a black dashed line. An asterisk indicates a significant difference to young at the peak ($p<0.0005$). All figures in this chapter show ± one standard deviation as areas around each profile, where grey = elderly and blue = young

A reduction in shoulder width may lead to an increase in medio-lateral force on the arm rests. The further the shoulder joint is from the arm rests, the more adduction

moment would be required to push the body upwards, as shown in Figure 8.2. Here, the elderly have their shoulders in a more abducted position to account for the increased distance between the GH joint and arm rests at the start of Phase 1. As both groups reach peak hand forces at approximately 35% of the motion, there is a brief portion of adduction (which appears larger in the elderly group) before standing. In both cases, there were significant differences between the young and elderly ($p<0.0005$ at the start of the motion, $p<0.05$ at peak hand force).

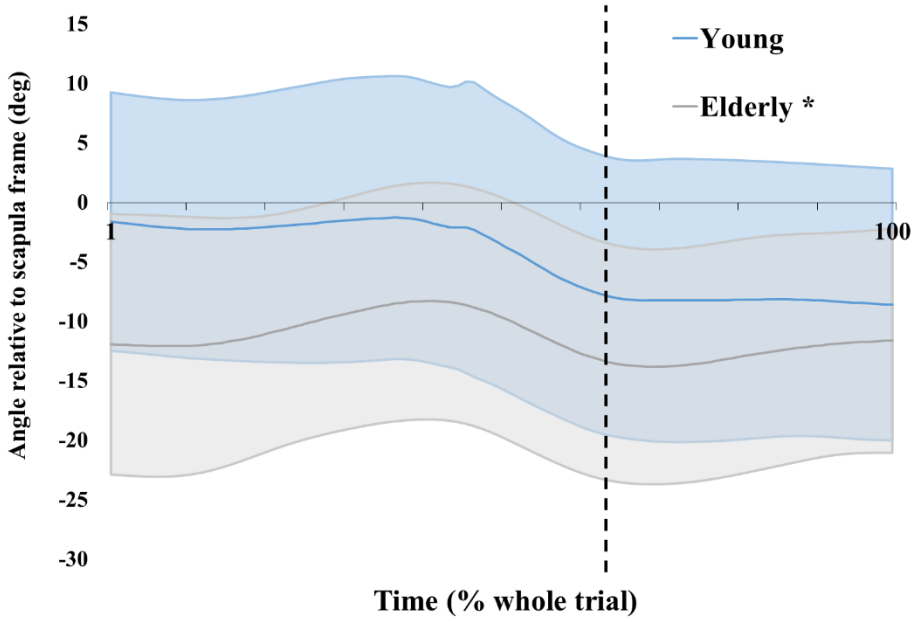


Figure 8.2 - Arm adduction whilst performing STS with the use of arm rests. Time at which the hands leave the arm rests is indicated by a black dashed line. An asterisk indicates a significant differences to the young at the start and at the application of peak hand force (~35% of the motion)

This increase in abduction prior to leaving the chair, and subsequent need to provide an adduction moment, may introduce medio-lateral forces on the arm rests. These were not measured in the current study, but the inclusion of these may result in increases in the calculated elderly GH JRF.

It is well-known that using the arm rests assists STS and is commonly recommended when treating elderly patients, as the upper body can assist in lifting the body without excessive use of the lower limb joints. However, it is evident that this can

place large loads on the upper body joints. In these cases, with extreme atrophy in the legs and discomfort or pain in the shoulder, the elderly will simply not be able to complete the STS activity without assistance. Many elderly patients are trained to rise without the use of arm rests to limit the dependence of the body on upper limb assistance. Therefore, difficulties in rising without arm support must be better understood.

8.2.1 Trunk and lower limb kinematics whilst rising without arm rests

Contrary to published data showing that elderly subjects have increased trunk flexion relative to the young (Alexander et al., 1991), the current study showed no significant differences between young and elderly peak extension values (Figure 8.3). The work by Alexander and colleagues was only for Phase 1 (time until the body leaves the seat) and only showed a 5-degree difference between groups. The absolute values of trunk flexion for the current study are in agreement with data from Anglin & Wyss (2000).

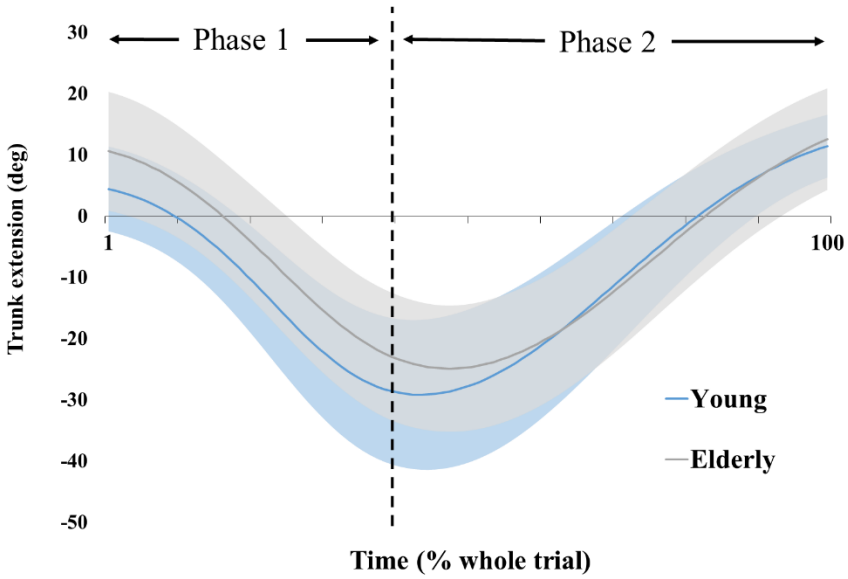


Figure 8.3 - Angle of the trunk relative to the global frame. Time of leaving the seat is indicated by a black dashed line. Phase indications apply to all following figures unless otherwise stated

Chapter 3 looked at the concept of 'postural stability', which describes an individual's ability to maintain their CoM over a base of support. This base of support would most likely refer to the feet, or possibly the centre of pressure. Published data

showed the elderly to require higher postural stability, and would therefore prioritise placement of CoM over the centre of pressure (CoP). However, in the current study this was found not to be the case when comparing age groups. All subjects reduced the distance between upper body CoM and foot CoP prior to leaving the seat, with no significant differences between groups.

In terms of postural stability, the current study is in disagreement with past literature on the priorities of STS who stated that increased trunk flexion was required (Alexander et al., 1991; Dolecka et al., 2015; Gross et al., 1998; Papa & Cappozzo, 2000). This may come as a result of the participants included in this thesis. The elderly group were mostly populated with subjects who regularly kept fit, were army veterans and who made use of in-house gym facilities. This may have led to an improved ability to stand up without excessive trunk flexion compared to those who featured in past investigations. Literature studies featured individuals who were: deliberately recruited due to known difficulties with STS (Alexander et al., 1991), not compared to younger control groups (Dolecka et al., 2015) or not shown to be statistically different to a younger group (Gross et al., 1998). There is therefore a distinct lack of evidence to suggest an able-bodied group of healthy older individuals would require an increased trunk flexion to succeed in STS.

As all subjects were asked to begin each trial with their knees and backs at approximately 90°, statistical tests revealed nothing of interest in terms of angular displacement. However, movement at the ankle was not so consistent between groups. There were significant decreases in the peak dorsiflexion angle (defined as negative in Figure 8.4) for the elderly compared to the young group. The relevance of this is further discussed in the next section.

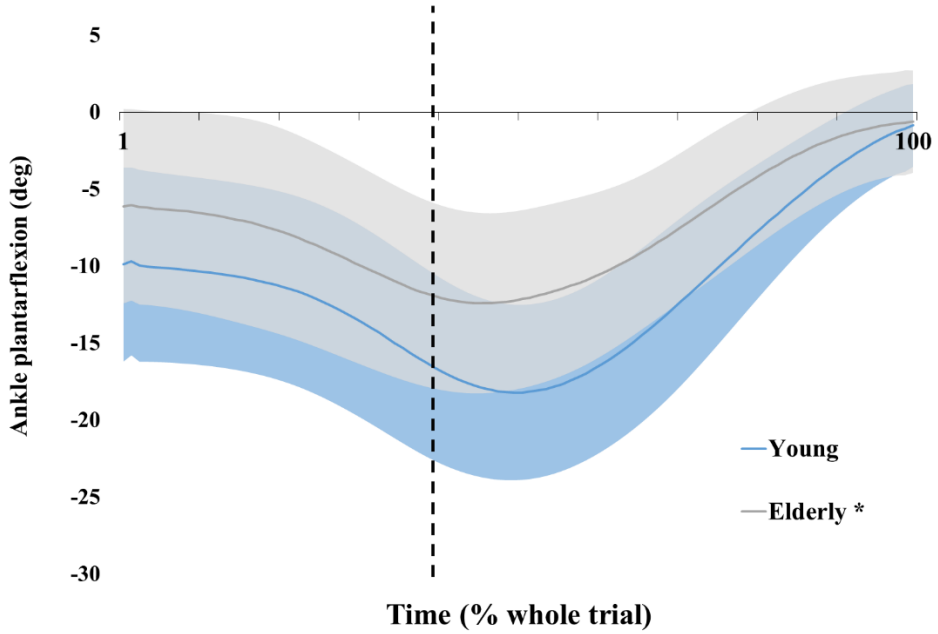


Figure 8.4 - Angle of the shank relative to the foot. Negative values represent dorsiflexion of the ankle. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young ($p<0.0005$) at peak dorsiflexion

Previous studies have shown increases in both knee and hip angular velocity with age (Schenkman et al., 1996), however, no significance was reported. The current study also found no statistical difference between the peak values of the knee and hip angular velocity (approximately mid-way through Phase 2) between young (knee – 122.5 ± 26.5 °/sec, hip – 143.7 ± 24.2 °/sec) and elderly (knee - 109.7 ± 28.4 °/sec, hip - 132.9 ± 41.1 °/sec), despite absolute differences between the two groups (Figure 8.5 and Figure 8.6). However, the group used in this thesis were typically slower in both hip and knee angular velocity in contrast to the literature (Young knee – 139.4 ± 18.4 °/sec, hip – 152.4 ± 20.1 °/sec and Elderly knee – 159.4 ± 33.5 °/sec, hip – 169.9 ± 46.5 °/sec - Schenkman et al., 1996). The study by Schenkman et al. included community-dwelling elderly participants, none of whom had joint replacements or reported arthritic diseases. This will have impacted upon the results, as 19 out of the 35 elderly participants in the current study had replacement knees and/or hips or stated they had joint pain.

During data collection, each participant was asked to give a brief medical history detailing any past joint surgeries or current joint pain. By removing the healthy (no joint

surgery or pain, n = 16) half of the group, the new value for knee angular velocity for the elderly was 103.4 ± 28.9 °/sec. This was found to be significantly different ($p<0.05$) to the knee angular velocity of the young group. The absolute effects of removing the non-surgery half of the group was evidently small, however removing the more extreme velocities of the healthy half of the group had a significant effect. All further analysis was performed on the entire group, but the statistical effects of including healthy and surgery-related subjects should be noted. Much the same was seen at the hip; the groups were generally not significantly different (Figure 8.6) in hip angular velocity at the peak of movement mid-way through Phase 2.

The similarities in velocity profiles explains the lack of significant differences between STS durations despite evidence to the contrary in the literature. The angular velocity of the ankle did, however, present a significant difference between young and elderly groups (Figure 8.7). The differences here, and the distances between ankle and centre of pressure data, may explain how the movement strategy has changed considering the drop in quadriceps activity seen in Chapter 7. This is further explored in the next section.

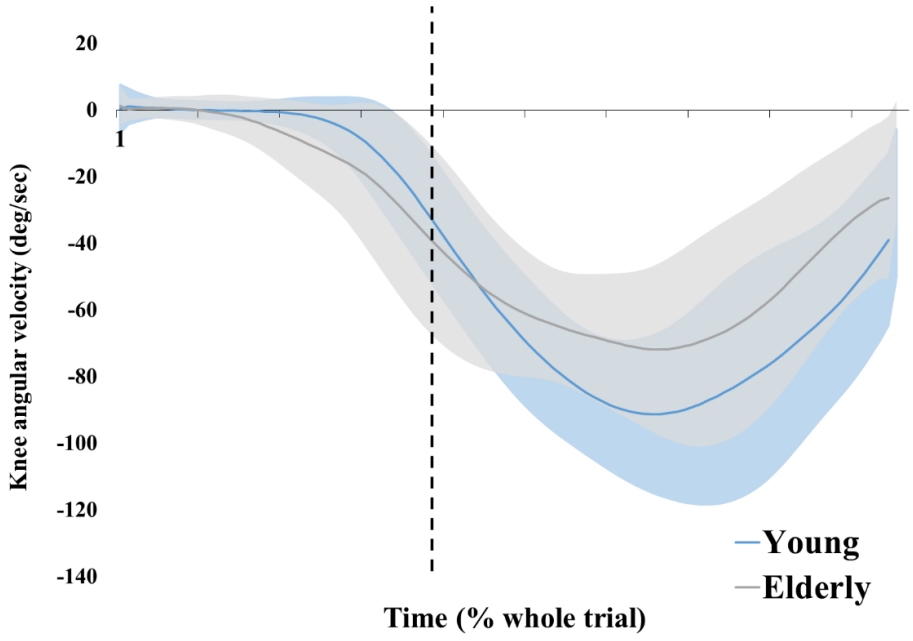


Figure 8.5 – Angular velocity of the knee joint. Negative values represent extension. Time of leaving the seat is indicated by a black dashed line.

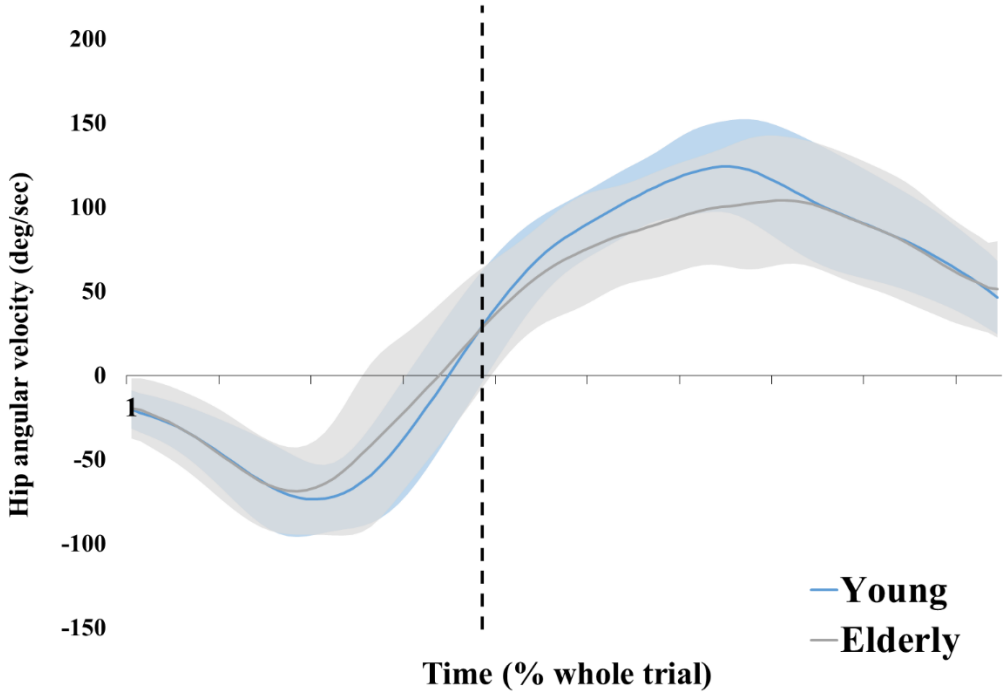


Figure 8.6 - Angular velocity of the hip joint. Negative values represent flexion. Time of leaving the seat is indicated by a black dashed line.

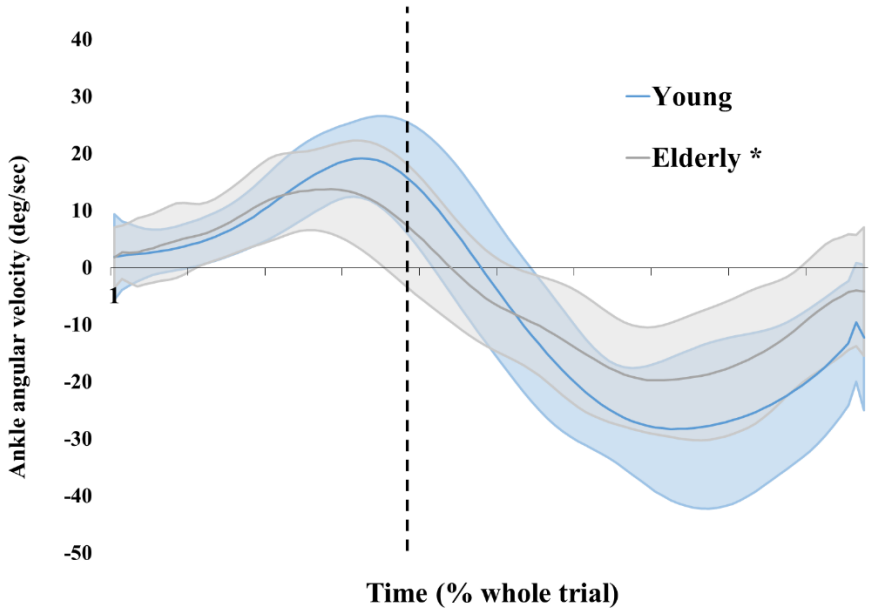


Figure 8.7 -Angular velocity of the ankle joint. Negative values represent plantarflexion. Time of leaving the seat is indicated by a black dashed line. Asterisk indicates a significant difference ($p<0.05$) to young at the minimum trough

The intersection of velocity profiles towards the start of Phase 2 for young and elderly groups suggests differing profiles in both knee and hip angular velocities. Further investigation revealed absolute differences in the acceleration of both joints at the point at which the subjects left the seat. However, these were not found to be statistically significant (Table 8.2). The crossing of the profiles, and increase in knee velocity at peak in the young, suggest in both cases that the elderly began hip and knee angular movements earlier. This may explain the significant differences in the kinematics, where a lack of statistical difference was found in Chapter 7 for durations of STS.

Table 8.2 – Peak accelerations for all lower limb joints.

Age group	Ankle (°/sec^2 ± SD)	Knee (°/sec^2 ± SD)	Hip (°/sec^2 ± SD)
Young	2286.1 (943.3)	7394.6 (2398.3)	8598.6 (2277.2)
Elderly	1875.4 (854.3)	6608 (2895.9)	7941.4 (3666.3)

8.3 Joint torque requirements

8.3.1 At the ankle

The elderly were shown, in Chapter 7, to produce significantly greater soleus and gastrocnemius muscle forces when compared to the younger groups. The kinematics and associated joint torques will be further explored to gain an understanding of why the elderly may be changing the ankle muscle recruitment strategy. Many elderly patients are advised, or naturally prefer, to move their feet backwards prior to standing up. Differences between GRF centre of pressure (CoP) and ankle for each group may come as a result of this, as shown in Figure 8.8.

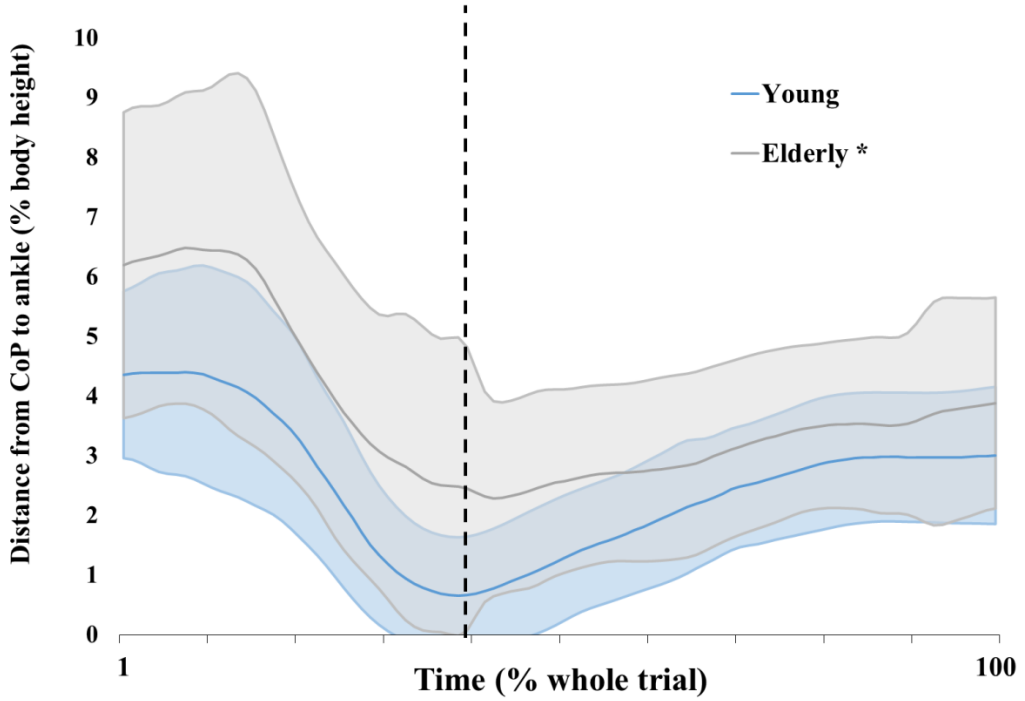


Figure 8.8 - Distance between the CoP of the GRF and the ankle joint. A positive difference indicates the CoP was anterior of the ankle. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young (p<0.0005) at the minimum trough

The elderly have placed their ankle further away from the CoP when compared to the young. Calculating the mean external moments at the trough (the point at which participants left the seat) by multiplying this distance by peak GRF:

$$YO: 0.7\ \%\ BH = 0.013\ m * 61.2\ \%\ body\ weight\ (peak\ GRF) = 0.013\ m * 481\ N$$
$$= 6.1\ Nm$$

$$EL: 2.3\ \%\ BH = 0.04\ m * 62.3\ \%\ body\ weight\ (peak\ GRF) = 0.04\ m * 520\ N$$
$$= 20.8\ Nm$$

The elderly group were therefore found to have a required moment at the ankle equal to three times that of the young group. This will undoubtedly affect the muscle forces required in order to counteract the external force. The required ankle torque is met by the summation of plantarflexion and dorsiflexion activity:

$$Mankle = PF * moment\ arms - DF * moment\ arms$$

Therefore, an increase in calf muscle activity would be needed to create the increased plantarflexion torque required in overcoming the external force in the elderly. Muscle force data from the MSK models revealed a significant increase in plantarflexor activity, from the young to the elderly, as a direct result of this (Figure 8.9). After leaving the seat, the centre of pressure of the GRF moves further from the ankle, continually increasing the ankle joint torque as shown by Figure 8.10. The previously shown plantarflexor activity, in all groups, acts to counter this, however, the increased distance between ankle and CoP for the elderly group results in a significantly larger net torque than that of the young (Figure 8.10) around the mid-portion of Phase 2. Significant increases in extensor digitorum longus, extensor hallus longus and tibialis anterior (dorsiflexors of the ankle, see Chapter 7) in the young means the required torque is temporarily close to 0 prior to seat off.

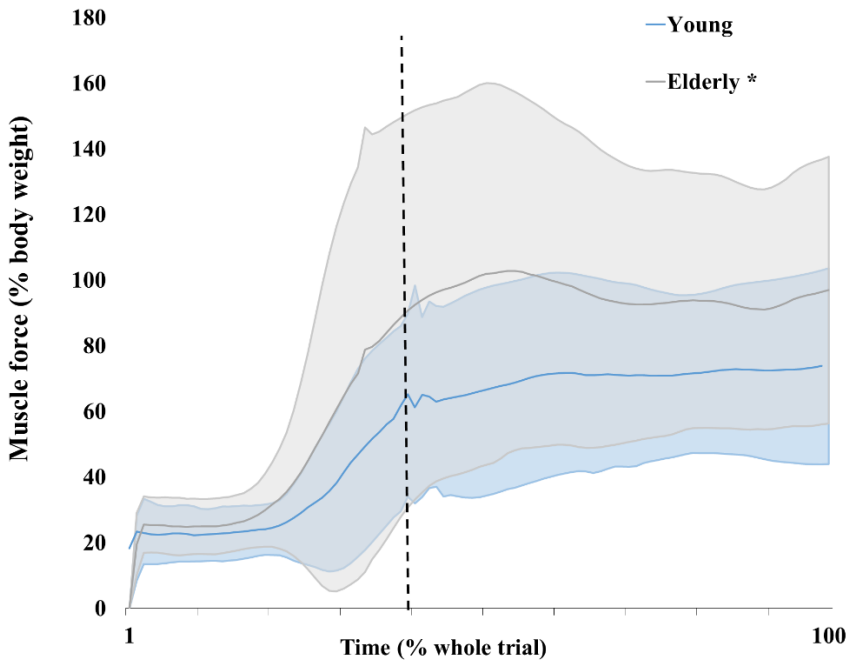


Figure 8.9 - Summation of all plantarflexor muscle activity when not using arm rests. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young ($p<0.05$) at peak.

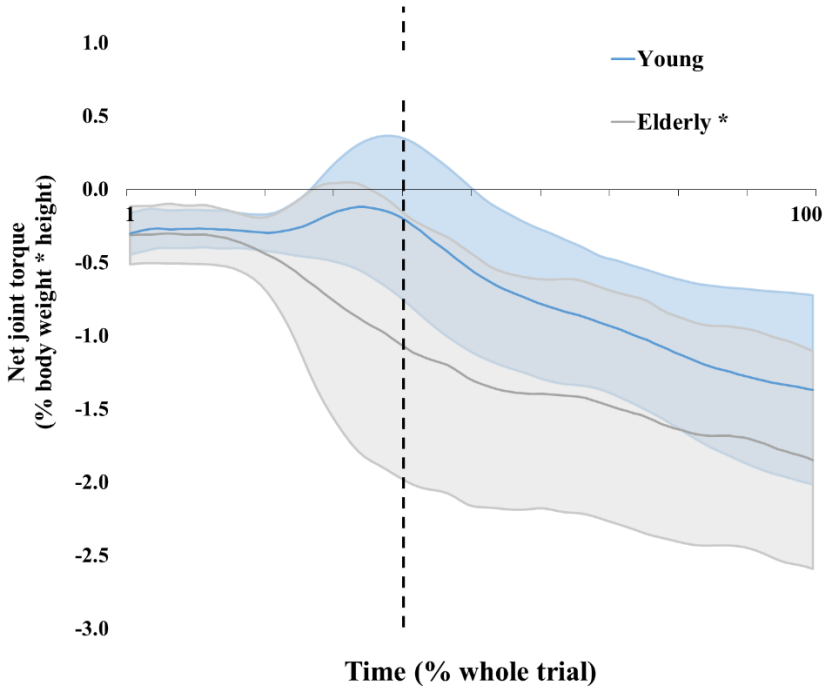


Figure 8.10 - Net torque of the ankle on the foot segment. Negative values represent an overall plantarflexion torque. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to the young ($p<0.0005$) midway through Phase 2

Joint power, which is the product of joint angular velocity and net torque (Figure 8.11), may explain differences in the strategies used by each age group in completing the STS activity.

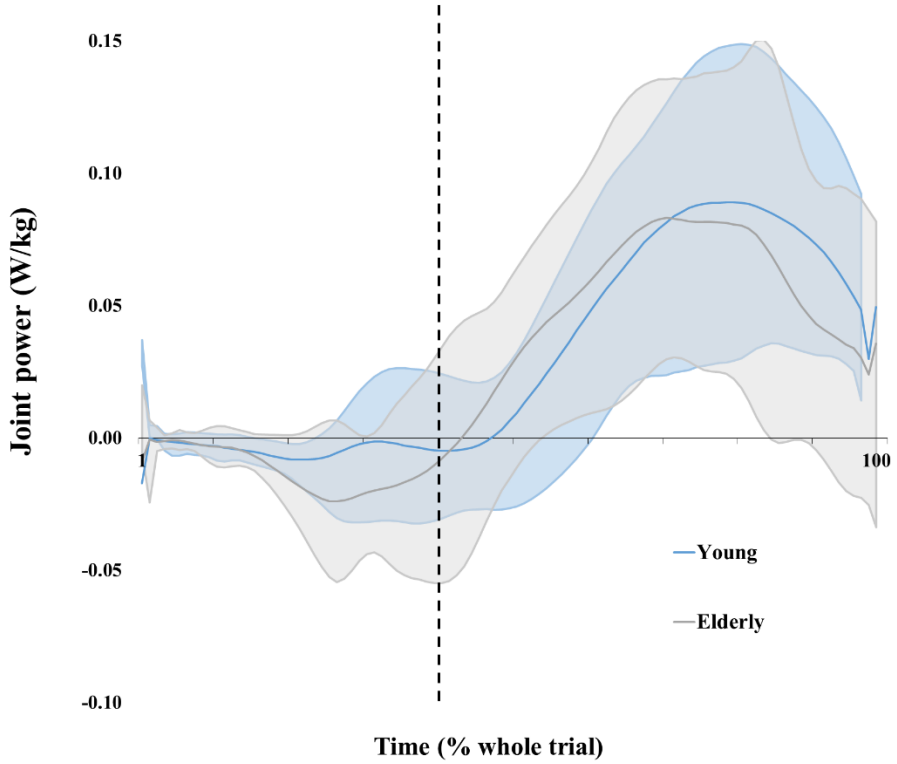


Figure 8.11 - Ankle joint power normalised to body mass. Time of leaving the seat is indicated by a black dashed line

The young and elderly are evidently exerting similar amounts of power during stand, however, the contributions to this power must be different considering the differences between angular velocity and net torque seen previously. With this in mind, it becomes clear that the overall contributor to joint power in the young is the acceleration of the joint as opposed to the elderly who contribute more net torque. Higher net torque is a result of increased muscle force contribution, and higher velocity is the result of increased acceleration.

Therefore, at the ankle, the elderly are relying more heavily on plantarflexion activity in order to stand up. This matches well with published data on elderly preference for stable movements over minimising required torques. Next, similar data for the knee will be explored to see if a similar theory applies.

8.3.2 At the knee

Chapter 7 showed the young group to have significantly, and substantially, larger quadriceps activity than the elderly group. This will have a profound effect on the JRF of the knee, in particular the patello-femoral joint which the quadriceps insert onto. As with the ankle, the kinematics and associated joint torque requirements will be explored. A consequence of the adjustment in CoP position in the elderly means it is generally placed closer to the knee. Therefore, when standing, the distance between CoP and knee joint is minimised (Figure 8.12).

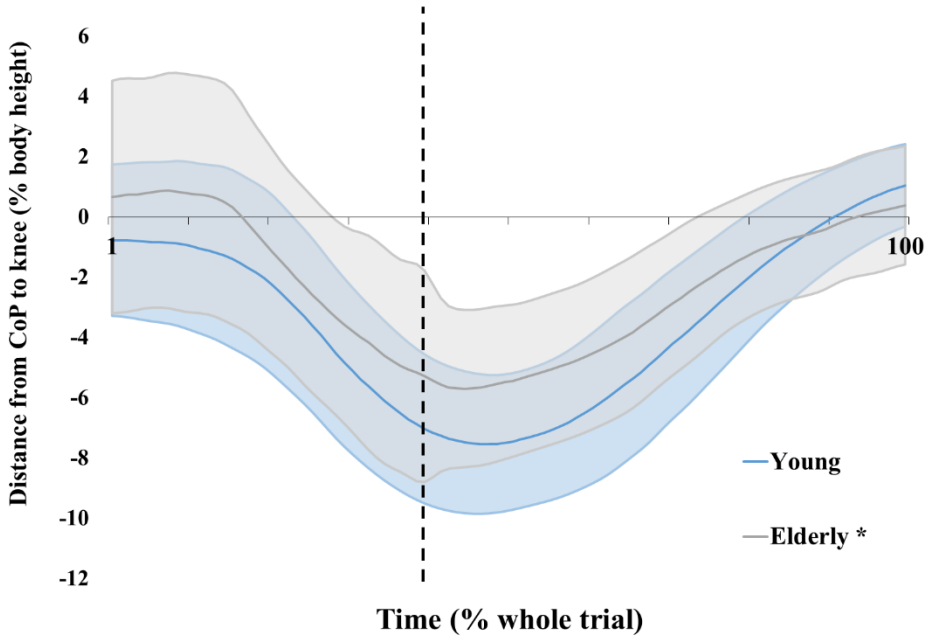


Figure 8.12 - Distance between the CoP of the GRF and the knee joint. A negative distance indicates the CoP was posterior to the joint. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young (p<0.05) at the minimum trough

This peak distance between knee and CoP will theoretically create a larger required-flexion moment in the young group:

$$YO: 7.5\ \%\ BH = 0.135\ m * 61.2\ \%\ body\ weight\ (peak\ GRF) = 0.135\ m * 481\ N = 64.9\ Nm$$

$$EL: 5.5\ \%\ BH = 0.095\ m * 62.3\ \ body\ weight\ (peak\ GRF) = 0.095\ m * 520\ N = 49.4\ Nm$$

The effect is not as large as that seen at the ankle, but is well matched with the clear reduction in the summation of knee extensor muscle activity (Figure 8.13) whose peak of activity is used to overcome this external flexion moment and lift the body over the knee during Phase 2. The summation of knee extensor activity (quadriceps) was significantly smaller in the elderly ($p<0.0005$) group when compared to the young group data.

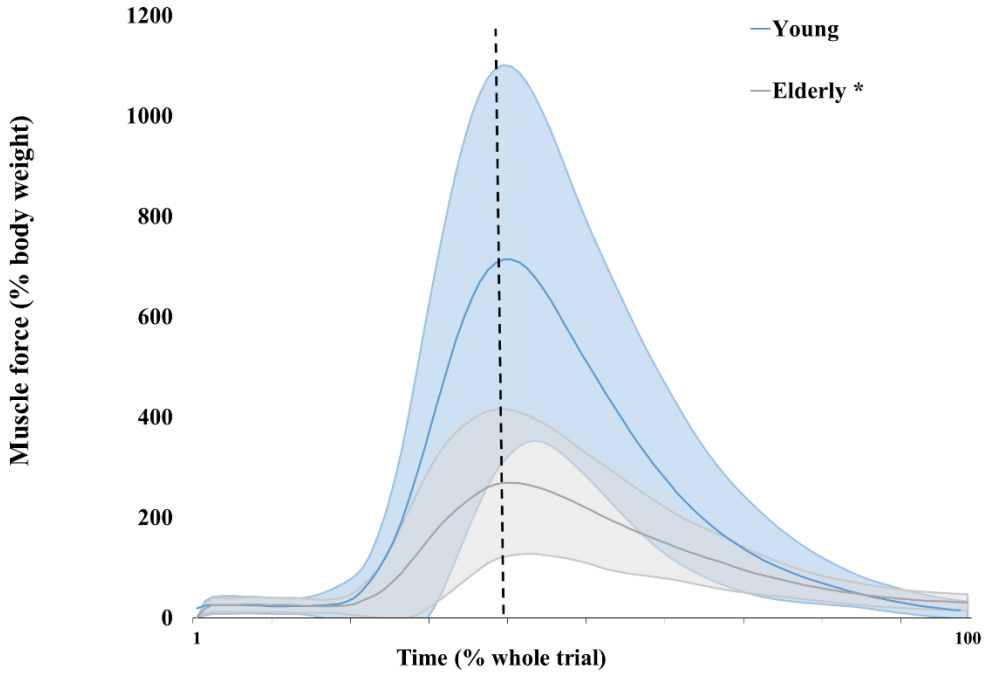


Figure 8.13 - Summation of knee extensor activity. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young ($p<0.0005$) at peak

This had a very clear effect on the net torque at the knee, where the elderly group had a significantly higher value mid-way through Phase 2 than the young group (Figure 8.14). As with the young group's data on the ankle just prior to leaving the seat, the knee moment reduces. This may be a result of the short peak of semitendinosus activity seen in the younger groups in Chapter 7. As a result, the net moment of the joint is larger in the elderly. This can be shown, as with the ankle, via the equations of motion applied during the inverse dynamics applied in MSK modelling. The equations differ for the shank, as acceleration of the joint, net torque at the ankle and distal contact forces must also be considered:

$$Mknee + Mankle + Weight * CoM - JRF * (distance\ from\ ankle\ to\ knee) = I\alpha$$

As the ankle was in plantarflexion (negative) but is equal and opposite on the shank so is now positive, and that the shank weight will also have a positive moment on the knee (considering it to be behind the knee CoR), then this becomes:

$$Mknee = I\alpha - (Mankle + Weight * (knee\ to\ CoM) - JRF * (ankle\ to\ knee))$$

The segment weight contribution does not change throughout the motion, however the ankle moment was previously shown to gradually increase. Assuming a constant acceleration of the shank (or very little movement), then the knee extension torque will also gradually increase as a consequence of the ankle joint torque. This is confirmed in Figure 8.14. The torque at the knee can be subdivided into the action of the knee extensors and knee flexors. As the semitendinosus is bi-articular, and also acts as a knee flexor, then the peak of activity previously found in the young group will act to marginally decrease net torque at the knee. This could be in preparation for a large peak of quadriceps activity to accelerate the thigh over the knee, subsequently rising the body out of the chair.

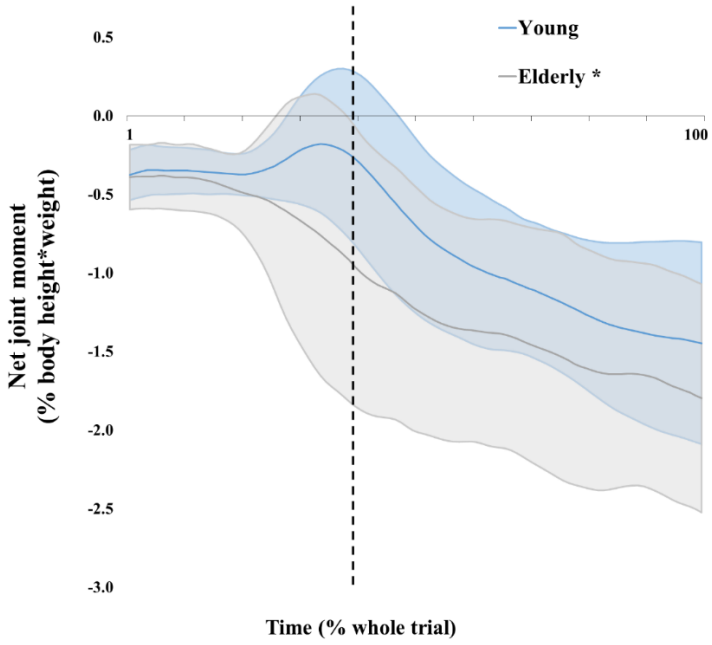


Figure 8.14 - Net torque of the knee acting on the shank segment. Negative values represent an extension torque. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young ($p<0.05$) mid-way through Phase 2

Joint power at the knee (Figure 8.15) was similar to that for the ankle in which there were no statistically significant differences between age groups, thus showing that contribution to joint power has shifted from angular velocity to muscle activity.

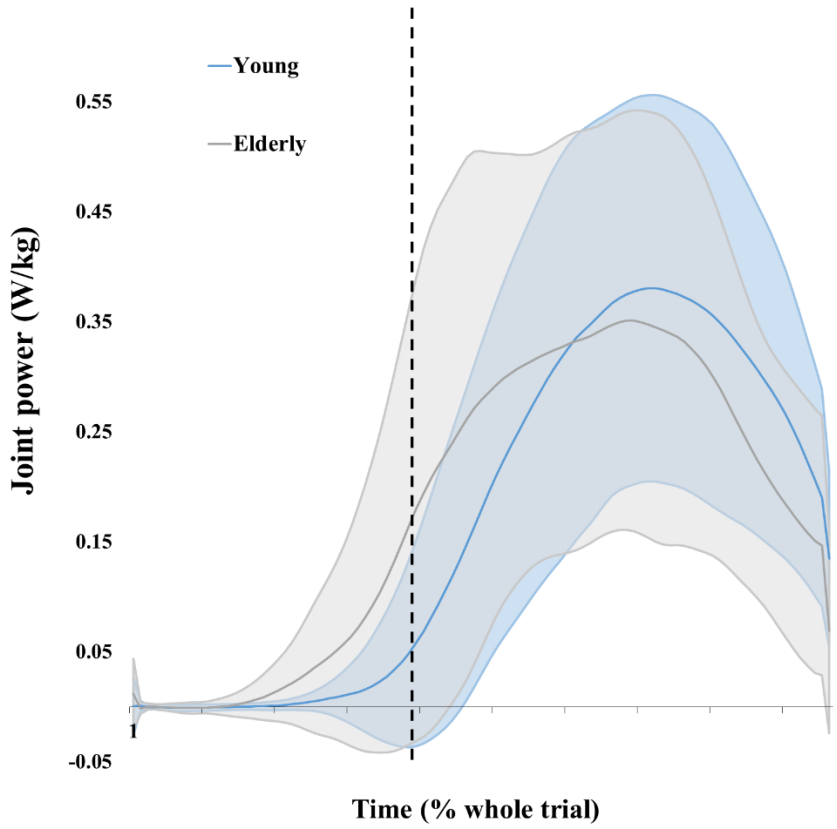


Figure 8.15 – Knee joint power normalised to body mass. Time of leaving the seat is indicated by a black dashed line.

In a similar fashion to the ankle, the elderly appear to be relying more heavily on muscle activity, as the quadriceps are still active despite a large reduction relative to the young, whereas the other groups have a higher knee angular velocity. This acceleration-related strategy is most likely a result of singular-activation of the quadriceps providing a large peak force just after leaving the seat. This would provide sufficient momentum to lift the body from a quasi-static state after leaving the seat to a strong, stable standing position.

The review in Chapter 3 on STS completion found that older groups favoured more stable movements, as opposed to meeting the required joint torques for efficient movement. The data presented in this thesis support this, with many of the muscle groups which act as antagonists to knee extension (knee flexors such as the gastrocnemius and the hamstrings) presenting with higher activity in the elderly group. Higher co-contraction leads to a more stable, albeit stiffer, joint. This would slow the movement but allow for a safer transition from sitting to standing.

From a qualitative point-of-view, the associated pain and discomfort in the elderly group, as shown by the large numbers of joint replacements or knee pain stated in given medical histories, would support the theory of joint protection. The reduction in quadriceps activity, and consequent reductions in both TF and PF JRF, shows that the elderly are avoiding use of this part of the leg. In light of known joint pain in an ageing population, the results presented in this thesis would suggest a much wider problem of avoiding the use of painful joints. In the young, they may be less susceptible to knee pain and can continue to move normally, whereas in the elderly this is evidently a much larger problem and a change in strategy has occurred.

8.3.3 At the hip

The elderly group were shown to significantly increase hip extensor activity, in the long head of biceps femoris and semimembranosus, relative to the young. This would suggest an increased contribution of the hip extensors to extension of the thigh, whereas in the young it is expected that momentum facilitates this movement. Similarly to both the ankle and knee, this will be explored below.

There were no significant differences in terms of angular velocity/acceleration between young and elderly. The profile for net torque is different than that seen for both the ankle and knee (Figure 8.16), in that it is the only lower limb joints to alter the direction of the net torque from flexion to extension during STS. Profiles were, however, similar amongst groups. As with the knee, the equations of motion governing the thigh may explain this:

$$Mhip + Mknee - Weight * CoM - KneeJRF * (distance\ from\ knee\ to\ hip) = I\alpha$$

Rearranging to find the torque at the hip:

$$Mhip = I\alpha - Mknee + Weight * CoM + KneeJRF * (distance\ from\ knee\ to\ hip)$$

At the start of Phase 1, knee JRF, acceleration and *Mknee* are all close to zero. Therefore, *Mhip* is predominantly balancing the weight of the thigh. As the moment at the knee (Figure 8.14) and acceleration (which will be in the negative direction) increase the moment at the hip will begin to decrease, the effect of which is greater than the knee JRF and thigh weight (owing to the reduction in sagittal-plane moment arm as the hip moves anteriorly during Phase 2), giving an overall negative (extension) hip torque. The continual increase in hip angular velocity (Figure 8.6) would suggest this joint relies more on earlier-built momentum for the later stages of Phase 2, however, as with both the knee and ankle this appears reduced in the elderly group. Instead, significantly larger hip extensor activity acts to lift the body.

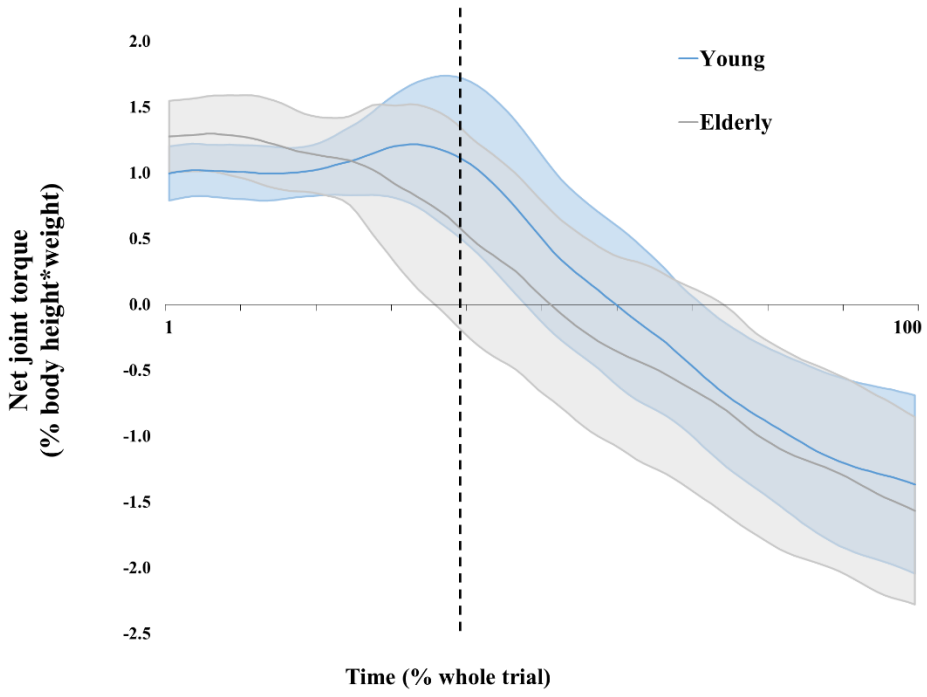


Figure 8.16 - Net torque of the hip acting on the thigh segment. Negative values represent an overall extension torque. Time of leaving the seat is indicated by a black dashed line

The significantly larger rectus femoris activity seen in Chapter 7 in the young group, around the point of leaving the chair, may well explain the small peak in hip flexion torque. The above graph also suggests that exertion of the hip extensors began earlier in the elderly but reached similar levels of torque as the young midway through Phase 2. Looking at hip joint power further reveals the effects of this (marginally but not statistically significant) earlier transition from hip flexion to extension (Figure 8.17). All groups begin with exertion of the hip flexors (or trunk flexors, which are beyond the scope of this thesis and not included in the MSK models used) to facilitate a lean forward. The net torque then suggests an early period of Phase 2 where there is still a flexion torque, but with an extensor angular velocity, and therefore a small portion of power absorption occurs before exertion of the hip extensors creates a positive joint power. The elderly, however, were found to have a significantly greater joint power than the young ($p<0.05$) at a point just after leaving the seat. This suggests the elderly begin exertion of the hip extensors earlier, and utilise extension of the hip for the majority of Phase 2. The power absorption of the younger groups may have synergised part of the hip extension, something which is evidently absent for the elderly. This supports the theory that the older group relied more heavily on muscle activity to drive hip extension, as with the other lower body joints.

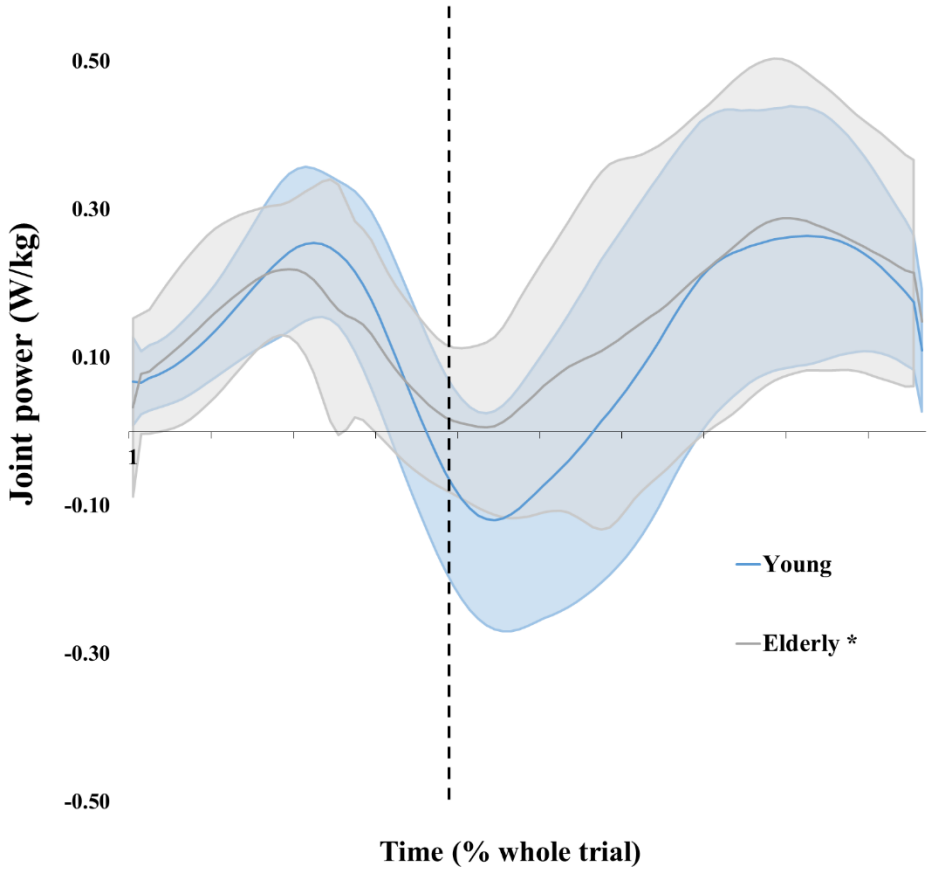


Figure 8.17 - Hip joint power normalised to body mass. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young ($p<0.05$) at the minimum trough

After leaving the seat, the summation of hip extensor activity was significantly higher in the elderly group compared to the young, with prolonged activity towards the end of Phase 2 (Figure 8.18). This led to an increase in impulse (defined as the area under a force-time curve, and describes a change in momentum). Table 8.3 shows the whole-trial impulse values for both hip and knee extensors for all age groups. The elderly group were significantly larger in terms of hip extensor impulse and significantly smaller for knee extensor impulse. Despite the young having approximately 2.7 times the peak knee extensor activity, the impulse values were only 1.9 times that of the elderly. This indicates the elderly may have been active for longer, increasing the long-term impulse. For the hip extensors this is almost certainly the case, as the difference in impulse does not match with the difference in peak muscle activity.

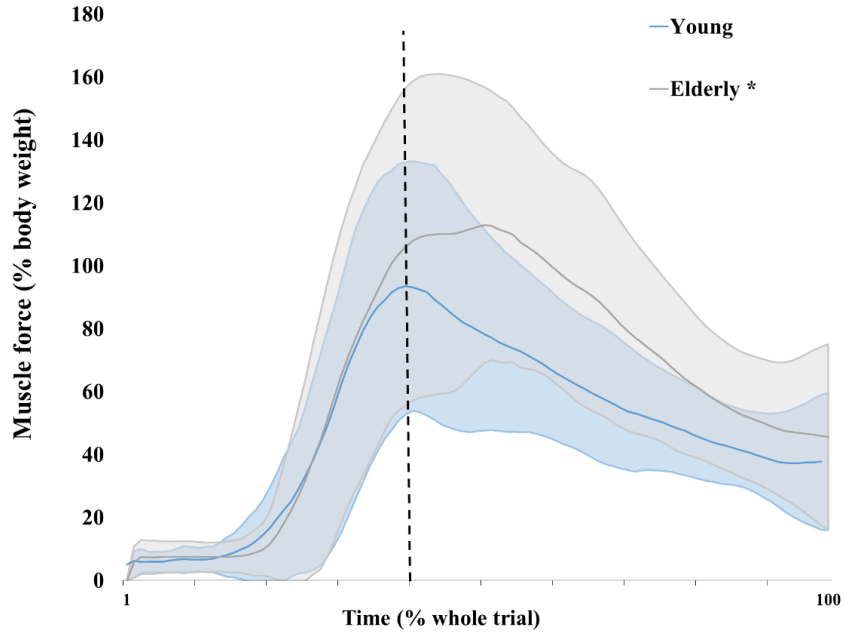


Figure 8.18 - Summation of hip extensor activity. Time of leaving the seat is indicated by a black dashed line. An asterisk indicates a significant difference to young ($p<0.05$) at peak

Table 8.3 – Whole-trial impulse values for all hip and knee extensors. [A] indicates a significant difference to the young group ($p<0.0005$)

Age group	Hip extensors (Ns ± SD)	Knee extensors (Ns ± SD)
Young	443.6 (167.3)	2949.2 (1646.3)
Elderly	769.9 (339.9) [A]	1570.7 (711.1) [A]

As momentum describes the motion of a body of mass (the product of velocity and mass), then these results further confirm that each age group was using different muscle groups to build momentum and move the body from sitting to standing, particularly during Phase 2.

8.3.4 Summary

It was found that the elderly moved their feet backwards to create a larger torque at the ankle in order to increase contribution of the plantarflexors to STS. Consequently, the required knee flexion was reduced in the elderly, so any quadriceps activity would purely extend the knee. A larger joint torque, but reduced angular velocity, shows the

changes in strategy at the knee to a more muscle-heavy movement. This was also seen at the hip, where prolonged muscle activity continually acted to raise the body up. Having investigated the associated movements for each age group, and concluded that a change of strategy is the cause behind changing strength levels, then it would be sensible to explore the different levels of muscle activation to explain how the effort levels of each age group are changing.

8.4 Muscle activation

Investigating the total activation of all muscles identified as producing significantly more force in the elderly group will help quantify differences in MSKR. Activation is defined as the produced muscle force relative to the MFUBs set prior to the loadsharing optimisation. Only muscle activations for the stand portion of STS have been presented in keeping with the rest of this chapter.

8.4.1 Upper body muscle activations

Table 8.4 shows the peak activation levels for all age groups. In many cases, the differences seen in this table will be due to the differences in applied MFUBs between age groups, which is confirmed by a lack of significance between some upper body peak muscle forces seen in Chapter 7. However, this table demonstrates that if the age groups are producing similar levels of absolute force, then the elderly will be working relatively harder considered the known reduction of strength with age. The elderly group had 4 muscles whose peak activation was greater than 50% activation, compared to 0 muscles in the young. These results suggest that completing a STS activity requires over half the available force-producing capability in certain muscles at peak effort for the elderly, and confirms previous work which found that this motion led to the elderly performing closer to their maximum capacity (Hughes et al., 1996).

As a general visualisation of muscle activation, Figure 8.19 shows the spread of data taken from the above table. Overall, the elderly are using more of their available strength to complete the activity, and therefore a reduction in MSKR is clear.

Table 8.4 – Peak upper body muscle activation when standing using arm rests. Full muscle names can be found in Chapter 6. Asterisks indicate muscles whose peak forces during stand were found to be significantly different between young * or middle-aged ** and elderly

	Mean peak activation (% muscle force upper boundary ± SD)		
Muscle	Young	Middle-Aged	Elderly
Trap_clav	5.2 (4.3)	7.5 (6.6)	7.2 (4.8)
Trap_scap	8.9 (7.7)	7.7 (5.9)	11.7 (8)
LS	10.4 (10.9)	10.2 (9.5)	18.8 (17.2)
Rmin	8.1 (7.3)	5.8 (6)	11.9 (11.1)
Rmaj	4.0 (4)	3.9 (4.3)	6.9 (5.5)
SA	19.4 (16.9)	21.2 (11.7)	31.8 (17.2)
Pmin *	2.9 (2.5)	4.1 (3.9)	7.3 (5.1)
LD	1.7 (3.1)	1.9 (3.2)	2.2 (2.4)
Pmaj_clav	2.9 (3.3)	2.2 (2.6)	4.8 (4.7)
Pmaj_thor	13.1 (8)	11.4 (7.9)	18.2 (12.2)
DeltA	15.1 (9.9)	10.6 (13.6)	18.5 (16.1)
DeltM	29.0 (21.0)	37.5 (27.6)	61.5 (27.6)
DeltP	10.9 (9.0)	14.4 (6.1)	19.5 (8.7)
SS *	19.2 (17.4)	31.6 (26.7)	50.5 (34)
IS	1.8 (3.4)	1.7 (2.5)	1.8 (2.6)
SBS	1.7 (2.6)	3.5 (5.9)	3.8 (9.8)
Tmaj *	0.2 (0.4)	1.2 (3.1)	0.5 (1)
Tmin	18.6 (15.2)	31.4 (21.2)	50.9 (21.1)
CB	1.5 (2.7)	4.2 (8.3)	3.9 (8.9)
BIC_short	2.5 (2.9)	2.7 (6.3)	4.2 (5.9)
BIC_long **	23.9 (18.3)	25.1 (22)	54.8 (31.1)
TRI_long *	9.7 (10.2)	18.1 (14.3)	28.7 (20.2)
TRI_med	23.2 (19)	23.5 (15.3)	39.5 (20.7)
TRI_lat	13.7 (10.8)	14.7 (10.5)	25.6 (15.8)
Bra	2.5 (1.7)	2.9 (3)	2.5 (2.2)
Anc	13.1 (13.7)	13.7 (10.7)	27.9 (22.2)
BRD	1.5 (1.5)	1.5 (1.4)	1.4 (1.1)
SUP_hum	1.2 (1.6)	1.2 (1.4)	2.3 (2.8)
PT_hum	0.7 (0.7)	0.6 (0.5)	1.3 (1.2)

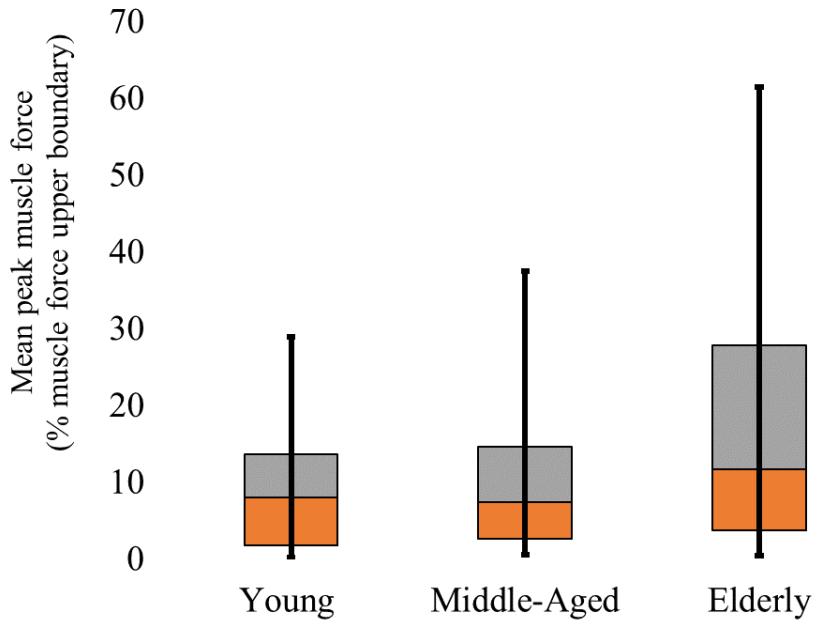


Figure 8.19 - Spread of mean peak muscle activation for all age groups during standing whilst using arm rests. Box plots contain data for all 29 upper limb muscles modelled in the UKNSM

Chapter 6 showed that some of the muscles in this table with high activations relative to the young, namely the serratus anterior, deltoids, triceps, biceps and teres major, contain a high proportion of fast-twitch muscle fibres. As the fast-twitch (Type 2) fibre is more prone to reductions in size and number with age, it becomes clear as to why the elderly may struggle to perform STS. If, with age, the strategies performed rely on the increased activation of these muscles then eventual age-related atrophy will hinder the success of the STS activity.

8.4.2 Lower body muscle activations

The peak muscle activations for the lower body are presented in Tables 8.5, 8.6 and 8.7, where both scenarios of with and without arm rests are presented. Considering the trials without the use of arm rests in the young, the muscles with large activation values are mostly the quadriceps (peak values of 73%, 59.7%, 81%, 61% and 50.3%) whose action propels the body over the knee to assume a standing position. The elderly were shown to alter their recruitment strategy, thereby changing the muscles with high activation levels. As a result, the number of muscles active over 50% are equal to 6 for both young and elderly groups. However, for the young these are 5 of the quadriceps muscles (the sixth being the semitendinosus). For the elderly, only 3 of the quadriceps are over 50% active. The semitendinosus, proximal sartorius and long head of the biceps femoris now feature peak activity over 50%. The spread of data for both with and without arm rests are shown in Figure 8.20 and Figure 8.21.

The quadriceps are known to contain a higher proportion of Type 2 muscle fibre (Brooke & Engel, 1969; Johnson et al., 1973; Manta et al., 1996; Ricoy et al., 1998; Staron et al., 2000; Garrett et al., 1984) than Type 1, so this muscle group can produce tetanic contractions at high velocities. Therefore, with the preferential loss of Type 2 muscle fibre with age, it would seem logical that an ageing body would not be able to produce such large quadriceps contractions over a small period of time. The inter-quartile range for the young is small with a median value around 15% (Figure 8.21). However, a large maximum value shows reliance on a small number of muscles.

The findings at the beginning of this chapter, and the clear changes in muscle activation levels seen in the tables below and Chapter 7, confirmed the alteration in movement strategy with age. Unable to produce fast, tetanic contractions mean the elderly must rely on prolonged activity of other muscle groups in order to lift the body. This increases the activation of muscles not normally active during STS.

Table 8.5 - Peak lower body muscle activation. Greyed data indicate muscles whose peak <u>forces</u> were found to be significantly different between age groups (see Chapter 7)

	Mean peak activation (% muscle force upper boundary ± SD)					
	Stand			Stand – no arm rests		
Muscle	YO	MA	EL	YO	MA	EL
Add. brev. (prox.)	4.7 (2.7)	7.2 (4.2)	8.0 (4.0)	10.7 (5.1)	15.1(9.1)	17 (10.6)
Add. brev. (mid.)	4.7 (2.7)	7.0 (4.2)	7.5 (3.8)	10.7 (4.9)	14.8 (8.8)	16.7(10.3)
Add. brev. (dist.)	4.7 (2.7)	6.9 (4.3)	7.3 (3.7)	10.8 (4.8)	14.6 (8.6)	16.5(10.1)
Add. long.	6.5 (3.7)	10.0 (5.9)	10.7 (5.2)	13.5 (5.2)	20.5(11.5)	23.5(13.3)
Add. magn. (dist.)	23.2(6.4)	25.2 (6.5)	29.8 (9.8)	33.3(10.9)	36.7 (8.8)	48.8(15.7)
Add. magn. (mid.)	14.9 (4.2)	16.6 (4.2)	19.5 (6.1)	21.6 (7.3)	24.8 (5.6)	32.8 (9.4)
Add. magn.(prox.)	8.2 (2.6)	10.2 (3.3)	11.2 (4.0)	14.5 (5.5)	17.7 (5.5)	22.6 (7.6)
Bic. fem. CL	14.0 (9.3)	21.5(10.8)	35.2(19.1)	15.6 (10)	32.9(22.9)	57.4(26.7)
Bic. fem. CB	4.7 (3.9)	6.6 (7)	9.1 (9)	4.9 (3.4)	7.9 (9.8)	9.4 (8.3)
Ext. dig. long.	12.8(14.2)	8.4(10.5)	8.6 (9.2)	20.3(22.8)	15.4(27.2)	10.5 (7.9)
Ext. hal. long.	13.3(14.8)	10.6(12.1)	9.1 (7.7)	25.3(26.4)	17.6(28.1)	10.0 (9.0)
Flex. dig. long.	1.6 (1.5)	3.7 (4.9)	4.4 (4.3)	8.4 (13.1)	8.4 (24.4)	5.9 (4.2)
Flex. hal. long.	2.2 (1.3)	4.2 (4.5)	6.2 (2.8)	10.1(15.8)	7.9 (18.4)	8.4 (6.3)
Gastrocn. (lat.)	8.7 (6.0)	12.3(12.1)	16.4(10.6)	9.5 (6.2)	12.7 (13.0)	17.2 (8.5)
Gastrocn. (med.)	4.5 (4.5)	7.8 (6.4)	16.6 (12)	5.0 (6.1)	6.3 (4.2)	16.3(13.5)
Gemellus (inf.)	7.7 (3.0)	10.5 (3.9)	12.3 (6.0)	14.4 (7.9)	21.1(23.9)	22.6 (7.4)
Gemellus (sup.)	6.8 (2.2)	9.6 (3.4)	11.4 (4.4)	12.0 (6.6)	17.8 (24.0)	19.2 (6.8)
Glut. max. (sup.)	12.0 (3.9)	17.9 (8.3)	19.7 (8.7)	16.3 (6.0)	23.8(10.3)	30.3(13.4)
Glut. max. (inf.)	6.0 (2.4)	6.9 (3.8)	10.6 (6.1)	8.5 (4.9)	9.5 (13.0)	13.2 (9.6)
Glut. med. (ant.)	6.4 (3.1)	8.8 (5.6)	12.1 (7.1)	7.0 (2.8)	9.0 (6.0)	12.1 (6.8)
Glut. med. (post.)	11.2 (3.8)	13 (5.0)	19.6 (8.8)	11.3 (3.9)	13.2 (4.7)	19.8 (7.9)
Glut. min. (ant.)	7.2 (3.7)	10.6 (7.5)	13.7 (7.7)	7.5 (3.3)	10.6 (7.8)	13.7 (8.1)
Glut. min. (mid.)	6.3 (2.3)	10.3 (5.6)	14.1 (5.1)	7.2 (2.7)	10.6 (6)	14.7 (5.1)

Table 8.6 - Peak lower body muscle activation. Greyed data indicate muscles whose peak <u>forces</u> were found to be significantly different between age groups (see Chapter 7)

	Mean peak activation (% muscle force upper boundary ± SD)					
	Stand			Stand – no arm rests		
Muscle	YO	MA	EL	YO	MA	EL
Glut. min. (post.)	5.7(2.2)	10.1(5.6)	14.7(6.1)	7.5(3.3)	12.1 (9.0)	15.1(6.7)
Gracilis	16.2 (8.0)	16.2 (7.3)	15.8 (6.2)	42.7(24.6)	31.2(12.7)	34.5(13.7)
Iliacus (lat.)	4.4 (2.6)	10.0 (5.1)	13.7(6.9)	6.8 (5.5)	11.4 (8.2)	13.0 (7.1)
Iliacus (mid.)	6.1 (3.7)	13.8(6.7)	19.1 (9.7)	8.1 (5.3)	15.9(11.5)	18.0(9.7)
Iliacus (med.)	4.7 (2.8)	10.7 (5.3)	14.6 (7.4)	7.0 (5.5)	12.3 (8.8)	13.9 (7.5)
Obt. ext. (inf.)	8.8 (4.1)	12.1 (5.5)	12.6 (6.3)	17.1 (8.6)	26.7(24.8)	28.4(13.9)
Obt. ext. (sup.)	13.6(6.8)	18.8 (9.1)	20.4(10.7)	25.7(12.7)	38.6(18.2)	44.1(17.7)
Obturator int.	16.3 (6.3)	21.6 (8.2)	24.5 (9.6)	28.3(12.9)	40.8(25.1)	48.4(16.9)
Pectineus	3.2 (1.8)	5.7 (2.6)	8.3 (5.1)	6.0 (5.5)	7.7 (6.3)	9.1 (5.1)
Peroneus brev.	13.7(12.6)	11.0 (12.2)	8.7 (9.3)	23.4(22.4)	16.6(27.2)	11.6(12.5)
Peroneus long.	5.8 (3.5)	8.8 (6.7)	10.2 (9.9)	13.5(14.7)	15.8(23.9)	9.1 (8.5)
Peroneus tert.	13.4(16.1)	11.8(17.1)	8.4 (9.4)	23.9(25.7)	17.6 (31.0)	10.8 (12.0)
Piriformis	7.8 (2.4)	10 (3.5)	13.6 (5.2)	9.2 (3.6)	10.9 (3.9)	16.2 (7.2)
Plantaris	2.3 (2.0)	3.8 (4.4)	4.9 (3.4)	2.8 (2.6)	4.1 (4.1)	5.8 (5.0)
Popliteus	8.3 (12.2)	7.1 (7.6)	9.3 (9.0)	16.3(24.5)	9.2 (11.9)	13.0 (18.0)
Psoas minor	7.3 (3.8)	11.6 (5.7)	12.9 (7.1)	15.1 (9.2)	25.8(25.1)	25.6(14.4)
Psoas major	7.5 (4.5)	16.6 (8.1)	23.4(11.9)	9.2 (5.5)	19.3(13.9)	22.0 (11.9)
Quadratis fem.	13.6 (4.7)	17.5 (6.0)	19.8 (7.2)	22.6 (9.2)	30.4 (9.6)	39.0 (13.8)
Rectus fem.	43.6(20.7)	37.9(16.2)	39.7(18.2)	73.1(28.7)	63.3(25.4)	62.6(22.4)

Table 8.7 - Peak lower body muscle activation. Greyed data indicate muscles whose peak <u>forces</u> were found to be significantly different between age groups (see Chapter 7)

	Mean peak activation (% muscle force upper boundary ± SD)					
	Stand			Stand		
Muscle	YO	MA	EL	YO	MA	EL
Sartorius (prox.)	11.8(11.2)	23.4(15.3)	28 (17.1)	18.8(16.2)	41.5 (25.0)	56.0 (23.1)
Sartorius (dist.)	16.5(13.3)	34.1(22.5)	44.1 (23.0)	24.8(26.8)	41.3(39.4)	42.9(23.9)
Semimembr.	11.0 (6.5)	18.0 (7.7)	27.6 (13.0)	12.4 (7.2)	24.6(14.5)	42.5(19.3)
Semitend.	32.1(18.1)	30.1(13.7)	33.0 (11.7)	59.9(34.4)	48.1(22.8)	56.7(17.1)
Soleus (med).	3.5 (2.9)	8.1 (6.4)	19.5 (18.0)	5.5 (4.4)	11.1 (8.2)	21.5(15.1)
Soleus (lat.)	10.0 (5.1)	11.7 (8.2)	14.7 (8.4)	10.8 (5.7)	11.6 (8.8)	19.5(12.6)
Tensor fasc. l.	11.5 (7.4)	30.3(20.1)	38.8(19.7)	12.2 (9.0)	38.0 (38.5)	36.1(21.4)
Tibialis ant.	12.4(11.2)	21.3(16.8)	20.5(18.4)	25.8(23.4)	32.3 (23.0)	20.7 (18.0)
Tibial post. (med.)	2.2 (1.8)	6.2 (6.1)	10.9(12.3)	5.3 (6.1)	13.9(26.4)	12.8(11.7)
Tibial post. (lat.)	2.2 (1.8)	6.2 (6.1)	10.9(12.3)	5.3 (6.1)	13.9(26.4)	12.8(11.7)
Vastus interm.	29.6(14.2)	25.3(10.3)	24.9(12.8)	59.7(28.2)	41.7(17.8)	40.1(22.5)
Vastus lat. (inf.)	11.9(5.9)	11.7 (4.2)	12.0 (5.4)	27.6(18.8)	18.1(6.2)	20.2(11.4)
Vastus lat. (sup.)	58.3(22.8)	48.7(19.4)	49.1(23.6)	81.0 (24.7)	72.6 (32.0)	68.7(32.5)
Vastus med. (inf.)	16.7 (9.4)	20.6 (7.7)	21.5 (8.5)	27.3(18.7)	32.1 (11.0)	35.7(14.7)
Vastus med.(mid.)	32.8(14.6)	33.2(11.2)	34.3(14.4)	61.0 (26.6)	53.2(17.1)	57.7(22.5)
Vastus med. (sup.)	23.8(10.5)	21.3 (7.4)	22.1 (9.2)	50.3(24.4)	35.3(12.7)	38.9(17.1)

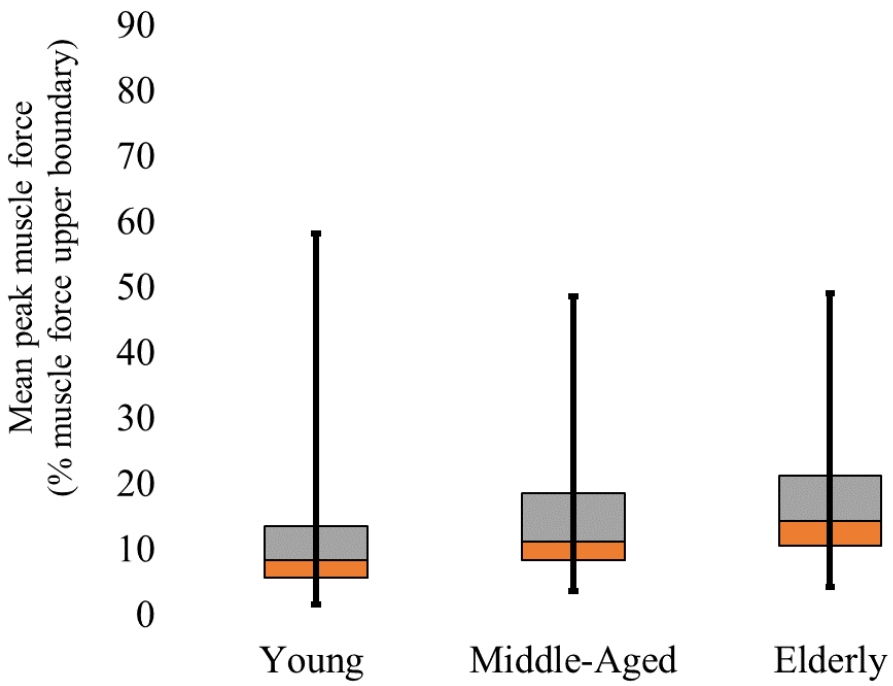


Figure 8.20 - Spread of mean peak muscle activations for all age groups during standing whilst using arm rests. Box plots contain data for all 58 lower limb muscles modelled in FreeBody

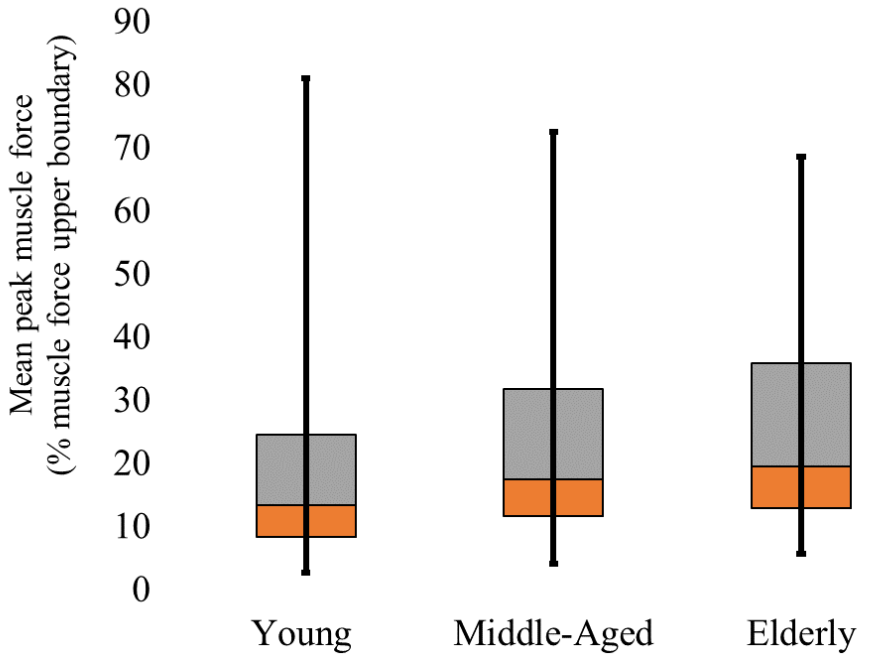


Figure 8.21 - Spread of mean peak muscle activations for all age groups during standing whilst not using arm rests. Box plots contain data for all 58 lower limb muscles modelled in FreeBody

Both figures show that the elderly group have an overall increase in muscle activation for both with and without the use of arm rests. The young have larger maximum values, confirming the previous theory of singular reliance on quadriceps activity. All age groups increase overall activity when arm rests are not used, but the elderly also appear to have a greater increase in interquartile range (with arm rests = 10.6, without arm rests = 23, compared to 7.9 and 10.6 for the young respectively) suggesting that a larger number of muscles are active in completing STS.

8.5 The Middle-Aged group

Although not the focus of this work, the middle-aged group was consistently seen as a 'step' between the young and elderly. In some cases, such as shoulder and hip joint reaction forces, or net moments of the lower body, they were found not to be significantly different to the young but were when compared to the elderly. However, the knee extensor activity and patello-femoral force showed significant differences between all groups. This would suggest the beginning of a shift in strategy, this is highlighted in significant differences for angular velocity. The young and elderly saw no statistical differences, however the middle-aged were significantly faster in the hip and knee. This could indicate a compensation to further increase momentum to overcome the required torques of STS. A lack of significant difference in muscle activity in both the plantarflexors and hip extensors between the middle-aged and young/elderly would support this theory. It may be the case that muscular strength has started to diminish in the quadriceps, however sufficient strength is still available not to have to fully rely on the prolonged 'muscle activity-strategy' of the elderly.

8.6 Limitations on Muscle Forces and averaging data

Allocation of subject-specific (in the elderly) and part-group specific (in the middle-aged and young) muscle force upper boundaries will have affected the muscle activations presented in this thesis. A lack of significant differences between absolute values for muscle force may come across as relatively large between age groups when normalised to maximum boundaries (as seen in Figure 8.19, Figure 8.20 and Figure

8.21). This simulated, despite similar levels of force applied to the movement, the elderly putting relatively higher amounts of effort into ADL completion and demonstrating the altering levels of MSKR. Future research may look to further quantify the changing levels of muscle force upper bounds on a subject-specific level.

Data presented in Chapter 7 was statistically tested based on subject-specific peak values, the associated onsets of which may have differed between participants as well as groups. When averaging across a group to given a 0-100% (of the motion) profile, these peaks may not always align. Therefore, peak data may not match that of the full profiles presented in this chapter.

8.7 Concluding remarks

The findings for a shift in muscle group loading are similar to the theories presented by Van der Heijden et al. (2009), who surmised, based on joint accelerations and net torques, that a shift in movement strategy from a momentum to a stabilisation-based approach would unload the knee extensors in favour of using hip extensors and plantarflexors. However, the stabilisation strategy was characterised by increased trunk flexion which was not seen in this thesis despite the evidential change in recruitment strategy. Van der Heijden and colleague's work did not feature such a comprehensive review of STS and was unable to confirm the action of individual muscle groups. It was also flawed in that the kinematics for the 'ageing' conditions were simulated via muscle weakness through increased mass. Younger individuals, which were the sole group for Van der Heijden's work, may have adapted to the loading in a different manner than older subjects. The current study has been able to address these limitations, and is therefore able to link changes in torque to alterations in muscle activity.

The work of this thesis presents an alternative theory as to the difficulties of the elderly population in performing STS, and by extension could apply to other vital ADLs. The young were found to rely more on acceleration of the body via activation of a singular muscle group, in this case the knee extensors. The elderly appear to rely more on prolonged muscle activity. This supports the theory of a shift in muscle fibre phenotype from fast to slow twitch. This would facilitate slower, but more stable,

movement but at the cost of increased muscle use. The main findings of chapters 7 and 8 are shown in Figure 8.22, where the changes in peak PF force are shown between young and elderly, as well as the redistribution of peak muscle activity from the quadriceps to alternative muscle groups.

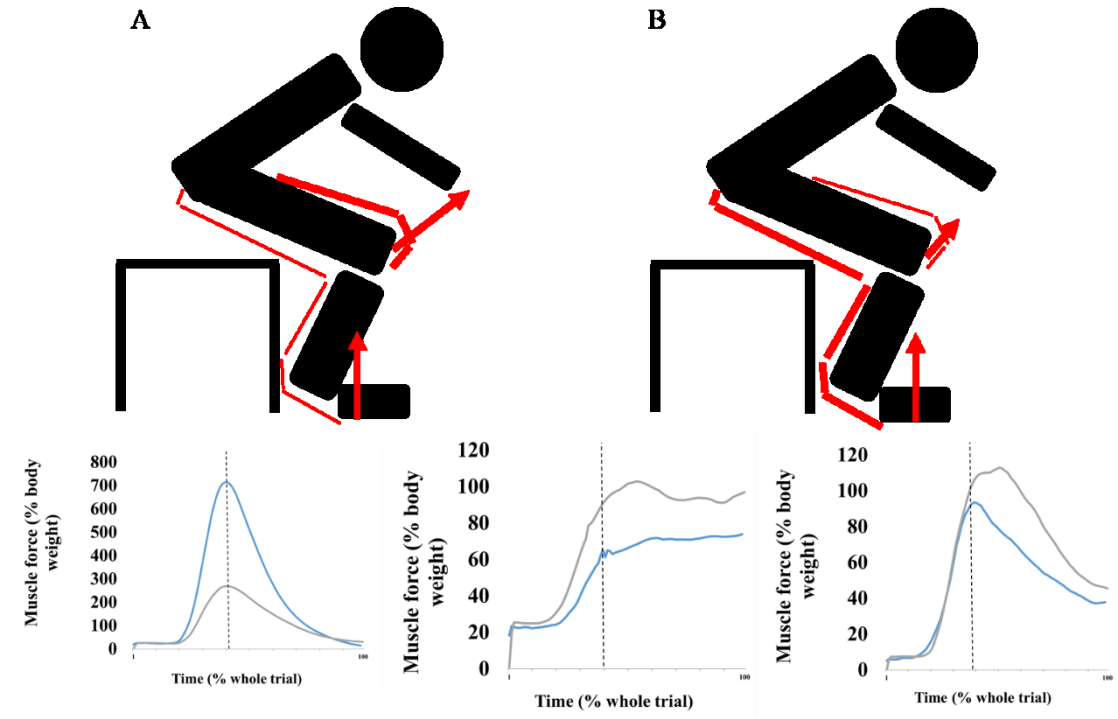


Figure 8.22 - Visual summary of the main findings in changes in muscle force (red lines – larger thickness indicates more active), CoP placement (lower red arrows) and PF JRF (upper red arrows – a larger arrow shows difference in magnitude) between the young (A) and elderly (B) groups during sit-to-stand at the point of having just left the seat. Graphs show differences in knee extensor activity (left), plantarflexor activity (centre) and hip extensor activity (right) between young (blue) and elderly (grey). These graphs are repeats of Figures 8.13, 8.9 and 8.18 respectively.

There are two possible definitions of MSKR. Firstly, as described in Chapter 3 in reference in exoskeleton design, MSKR could be defined as the difference between the task requirement and user strength/capability. This is sensible, as it considers the amount of assistance an individual requires in order to complete the activity. In the case of STS, this would mean providing enough quadriceps strength to overcome the required knee flexion moment and lift the body into an upright position. An exoskeleton would be designed to provide this assistance. This thesis proposes a new definition of MSKR. The data presented suggests that the elderly have shifted their recruitment strategy in order to accommodate a more stable, prolonged muscle activity-driven movement. This relies on new muscle groups, potentially the compensatory functions used by younger

groups when injured or weak. Therefore in the elderly, when the strength levels of the new muscle groups begin to fade and a natural redistribution of muscle activity is no longer possible, completion of the activity becomes almost impossible. MSKR could, in this instance, be describing the body's ability to respond to, and compensate for, a stressor event. This is the very definition of frailty described in Chapter 3. As the body begins to shut down, it can either adapt or suffer terminal effects on functional ability. Severely frail individuals were not featured in this thesis, mostly for safety reasons, but also because they most likely suffer from weaknesses which stop them being able to stand and sit unaided. It is possible to hypothesise, using the ageing data presented in this thesis, that continual weakening of elderly muscle will mean the new muscle-loading strategy will become a non-viable option. As MSKR is reduced to near 0 capacity, no further compensatory actions are available. This led, in an initial step, to using arm rests which have in-turn been shown to load the upper body at any age group but may be detrimental to the elderly in particular. There are two possible solutions to this: 1 - changing shoulder posture (or other kinematic changes) as shown through the clavicle rotational differences between groups and 2 – reactivation of the muscles which have seen significant reductions in muscle force. The body appears to be naturally redistributing muscle action as a result of localised weakness; in this case there is a noted redistribution in the quadriceps. It may be possible, through the use of functional electrical stimulation, to reactivate these muscles and restore a normalised pattern. When these become non-viable options, and muscle weakness reaches extreme levels, individuals are forced to remain seated and seek assistance from others.

The final chapter in this thesis will summarise the findings of chapters 7 and 8. The relevance to an ageing community will be discussed and future work pertaining to the application of the findings to a clinical setting will be explored.

CHAPTER 9

FINAL CONCLUSIONS AND FUTURE WORK

This chapter will summarise the work presented in Chapters 8 and 9, as well as the findings of the 3D scanning and muscle weakness chapters. All research is then placed into the broader context on how this work could be applied clinically and the relevance to an ageing population.

9.1 Concluding remarks

Musculoskeletal reserve is a term that is not generally used in the context of biomechanics, however, as defined in this thesis it provides a way of quantifying musculoskeletal functions. This is important in ageing, where a reduced physical reserve has detrimental effects on the body's ability to complete activities of daily living. Known reductions in muscular strength and ADL performance have been widely investigated, but the comprehensive biomechanical effects had not been explored prior to the work in this thesis. This work, through the acquisition of motion capture data and use of musculoskeletal modelling, assessed multiple age groups performing a key functional task associated with high strength requirements and difficulties in the elderly - the sit-to-stand activity - and determined the kinematics and kinetics of human movement.

It was initially found that, using young kinematics, weakness of the serratus anterior (and to a lesser extent, other fast-twitch dominant muscles) could prove detrimental to the success of an individual completing a STS activity when relying on upper body support. Weakness of the serratus anterior, a muscle important for scapula stabilisation, proved the most debilitating and greatly increased the activation level of remaining muscles. Known diseases relating to SA weakness are known to cause difficulties during functional activities, and this work not only confirms this, but highlights the ability of MSK modelling in the assessment of the effects of muscle weakness. Despite the fact that the SA does not cross either the glenohumeral or elbow joint, JRF at both sites were significantly increased when SA was weakened. This highlights the importance of understanding secondary, or even tertiary, effects of muscle weakness particularly in the context of ageing. Further knowledge of how atrophy affects movement may inform the clinical rehabilitation pathway on an individual basis.

A comprehensive study of three different age groups performing a STS activity revealed differences in how the upper and lower limb musculature adapts to changes in movement strategy. In the upper limbs, the elderly were shown to load the shoulder significantly more than the two younger groups which was found to potentially be a consequence of shoulder positioning. The reason for the change in kinematics is unclear, but the compensations manifested as an increase in shoulder muscle activity. Over-

reliance on arm rest support may accelerate atrophy of the lower limbs due to disuse, furthering STS difficulties.

In the lower limbs, the elderly appeared to be protecting the knee joint, in particular the patello-femoral joint, highlighted by significant decreases in both contact and quadriceps forces. However, this may also be interpreted as a result of quadriceps weakness rather than joint pain. In the young, the quadriceps were hypothesised to facilitate greater acceleration and therefore momentum was the greatest component of STS success. For the elderly group, in the absence of high levels of force in the knee extensors, prolonged muscle activity was the key to lifting the body to an upright position. This was confirmed through analysis of joint angular velocities and net moments. The young had significantly greater magnitudes in angular velocity, but the elderly provided larger joint moments. Considering that the joint powers (product of angular velocity and net moment) were the same, this suggested a change in strategy. Joint power, which is needed to move the body, was provided through different means, for example increased plantarflexor and hip extensor activity. This would have a large effect on the MSKR remaining in the elderly body, as more muscles are required to complete the activity leaving little capacity to compensate in the face of age- or injury-related muscle weakness. Quantifying this concept may further a clinician's understanding of how to improve ADL performance and may well change the way of approaching rehabilitation.

Previous studies have investigated the effects of age on postural stability during STS. The work in this thesis supports the theory of an increase in co-contraction in the elderly, as multiple antagonistic muscle groups were found to have greater activity during the motion. Even though there were no significant differences in CoM positioning between age groups, which is important when achieving a safe movement, this increase in muscle activity would suggest stiffening of the joint to improve postural stability. In terms of MSKR, and similarly to the discussions above, this reduces the potential reserves of compensatory muscle action, furthering the vulnerability of the elderly's movements in the face of continuing atrophy. This creates a cycle of detrimental effects on functional ability, in a similar fashion to that of frailty (as introduced in Figure 3.1). Age-related muscle weakness will naturally cause changes in movement strategy, as seen through both middle-aged and elderly groups. The work of

this thesis shows that this significantly increases the activity of compensatory groups of muscles, such as the posterior calf and thigh muscles. Eventually, when reliance on these muscles reaches extreme levels and MSKR is severely reduced, no further compensations are available and STS ability is extinguished.

Chapter 5 presented a successful method of producing subject-specific body segment parameters using 3D infra-red scanning, addressing disadvantages in previous methods for cost, speed and safety. The results were comparable to empirical data from cadaver-based studies and were more comprehensive than other studies presenting regression tables in terms of the number of segments and inclusion of inertial properties in multiple directions. The designation of relevant segment boundaries improves upon other 3D scanning methods and comparisons with both cadaver and geometric modelling data has not previously been presented in full. It is advised that future studies making use of MSK modelling consider implementing such measures to improve the specificity of the outputs.

9.2 Future work

This thesis proposed a new way to use 3D scanning to obtain body segment parameters. This technique, that provided data quickly, could be further developed to make the equipment safe for any age group, as the current iteration would not be suitable for someone unable to stand on the rotating platform independently. Future iterations of the design and processing process should address issues in capturing feet data and producing improved accuracy for head BSPs. Removal of hair data, or appropriate use of head-wear, would address the latter. Inclusion of a small ramped pedestal may improve the former. However, the material would need to allow for infra-red light to pass through and capture the full form of the feet. Development of an algorithm to automatically identify anatomical landmarks would decrease processing time and allow for almost instantaneous data acquisition in a motion capture/MSK modelling environment. Group-specific regression should also be investigated further: The results in this thesis showed that using gender and BMI as grouping variables provided the best approximation of BSPs using regression.

Chapter 6 presented a useful method of assessing the effects of muscle weakness on STS performance. It was found that the serratus anterior was pivotal in standing with the use of arm rests, however, this would need further investigation on participants with known serratus weakness. Scapula winging, a condition which can be caused my serratus palsy, may be one avenue worth exploring. Asking patients who suffer from this condition to perform STS just using their arms (no contribution from the legs) and observing the kinematics may reveal the extent to which this particular condition hinders movement. As standing using the arms requires a large amount of elbow extension, observing this planar movement may reveal changes in strategy, if complete reliance on arm rest support is assumed.

The overall findings of this thesis in terms of the elderly performing STS would be most useful in a clinical setting. Having observed a geriatric ward and consulted with geriatricians and physiotherapists, it was clear that geriatric wards encourage the elderly to stand and sit unassisted, however, many are trained to use arm rests or to increase trunk flexion. The effects of the former have been presented in this thesis (chapters 7 and 8), and the latter would theoretically increase the activity of plantarflexors and hip extensors owing to the anterior position of the CoM over the knee joint. This thesis has shown that introducing the activity of these muscles significantly increases the required joint torques, raising the overall task requirement for the elderly. However, an alternative solution would be to focus rehabilitation on the quadriceps. This group is mono-articular (barring rectus femoris); strengthening biarticular muscles (hamstrings, gastrocnemius) may result in secondary effects on the multiple joints they cross. Currently, clinical staff focus on building momentum, but the effects of this training technique are unknown. Instead, re-training the quadriceps would restore the strength required to pull the body over the knee and overcome the external knee flexion torque which seems to be difficult for the elderly. Increasing the proportion of fast-twitch muscles through strength conditioning may assist this. It is important to note that muscle recruitment is but one of a number of different methods through which functional ability can be improved. Despite the success of strength training regimes, clinicians must also address a patient's psychological barriers, diet, willingness to continue therapy and other non-physical elements of rehabilitation. The work in this thesis forms the physiological element of this, but does not inform other domains.

Inclusion of a middle-aged group provided a better understanding of when the changes exhibited between young and old occur. The inconsistency in differences between the middle-aged and others groups suggests that grouping by age should not be considered. Instead, age could be used as a continuous variable, whereby large groups of people are analysed and the trends over time are observed.

The methods applied to quantifying MSKR were only investigated for STS in this thesis, however, similar protocols could be implemented for other important everyday activities such as gait or ascending stairs. Future investigation could implement a similar methodology, whereby different age groups are observed performing a range of ADLs, and MSK modelling is used to assess how changing strength levels affects performance. Looking at the joint powers, and the individual variables of angular velocity and joint torque, would reveal the main contributors to movement. It would also reveal which muscles are working hardest to facilitate movement, and would further inform clinical intervention as with the above recommendation of quadriceps training.

Implementation of the findings, and these discussions, should constitute the next step in this research. However, the effectiveness of retraining (as opposed to a lifetime of regular exercise) a sedate population, such as the elderly, is relatively unknown. For an individual whose movements have gradually changed over time, as the middle-aged group data would suggest is happening at an earlier age than previously believed, then strength conditioning or training may not alter movement strategy but merely strengthen the muscles. Asking individuals to change long-term habits is difficult and could cause further problems. Instead, it may be more beneficial to train changes in movement or, as an alternative for younger patients, encourage regular exercise and strength training at an early age. If a group, such as those in the middle-aged bracket of this study, were aware of suitable training techniques or regular exercises to maintain everyday function, then the changes in strategy would most likely be ingrained into subconscious movement and may alleviate problems in the future.

The success of such work is subject to the MSK models used to produce the vast array of data. The limitations of current models mean improvements can be made, particularly in the validation of the calculated muscle and joint forces. EMG and

instrumented joint replacements are the current measures for validating MSK models, however, those with joint replacements are not indicative of a healthy individual. The same can be said for those undergoing invasive EMG techniques such as fine-wire, as discomfort will certainly result in altered kinematics. EMG is also notoriously difficult to direct correlate with muscle force. Therefore, improvements can be made in the validation techniques, to ensure that the MSK modelling results are accurate before using the data to inform clinical practice. Another area for improvement is the cost functions associated with the loadsharing optimisation when calculating muscle forces. The functions used in the UKNSM and FreeBody have been published and validated to a certain extent, however, no work has been done to show how these functions change with different scenarios. It is currently assumed that the muscle stress (UKNSM) or maximum muscle force (FreeBody) represent the body's priorities to protect muscular integrity, but it may be the case that these priorities change depending on injuries, age or function. Future work should investigate how these functions change, and the previous discussion on validation will also be required to confirm each function.

9.3 Broader Context

The work in this thesis has taken a biomechanical approach to addressing difficulties in the functional ability of an ageing population. However, there are other approaches available, including regular exercise, social interaction or improved national healthcare. A review by Cadore & Izquierdo (2013) found that participating in three training sessions a week, for up to 12 weeks, appeared to optimise strength gains in the elderly. However, the logistical nature of someone in their 80s or 90s participating in such high-demand training would be questionable. Overtraining was found not to occur, but functional outcomes may not justify the time commitments. Cardio-vascular performance is important in maintaining a high level of functional ability, particularly in the elderly. Endurance must be maintained, and concurrent training appears to improve this. The work presented in this thesis could inform this type of physiotherapy, or help create bespoke training regimes, as this form of training may not be suitable for everyone. The work of Chapter 5 may help to address this through improved MSK models of individual patients. The accuracy of the proposed 3D scanner means that body segment parameter inputs are more representative of the subject. This may prove

important when catering training regimes to different body types or, for example, through the identification of mass changes in individual segments which may indicate gains or losses in mass through training.

Society has changed in the last 100 years, as families provide less inter-generational support. Sander et al. (2014) discuss in detail the detrimental effects this can have an on elderly individual, and how social detachment can have significant health risks. Improving the body's ability to act independently will enable isolated individuals to interact socially. This may also address the every-increasing cost of treating the elderly, as more people are living longer and require round-the-clock care. Improving functional ability should restore independence in many elderly individuals, relieving the strain on the NHS and allowing individuals to remain part of a workforce for longer (if desirable).

The tools used to assess functional ability, disability and frailty were briefly discussed in Chapter 3. These tools are proven methods and are largely implemented in clinical practice, however, most come with a range of subjective questions and tasks. The methods used for this thesis would constitute an empirical tool for functional assessment in terms of producing muscle force exertion, muscle activity and forces through individual joints. This would help inform clinical practice through more objective measures, providing clinicians with a detailed understanding of their patient's body during functional, daily activities. Informed strategies would then form a key part of physiotherapy, as the data would prove more reliable than existing, qualitative questionnaires which can rely heavily on self-assessment.

The work in this thesis is the most detailed analysis of the sit to stand motion in the literature and has found clear changes from the young to the elderly that have pointed towards potential clinical interventions. The task was selected based on its importance for mobility and its association with social isolation. It would be conceivable to create a hierarchy of functional tasks, analyse them all, and then use the outputs to devise a holistic set of clinical interventions that could prioritise achieving these key functional tasks. Notwithstanding the complexity and difficulty in delivering a study that could produce such an output, it is the proposal here that this is urgently addressed as future work following on from this thesis.

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APPENDICES

Appendix 1 - Definition of segmental coordinate system in FreeBody

Anatomical landmarks used in coordinate system definition
MT1: Proximal 1st metatarsal
LM: Lateral malleolus
MM: Medial malleolus
LFE: Lateral femur epicondyle
MFE: Medial femur epicondyle
HJC: Hip joint centre of rotation
RASIS: Right anterior superior iliac spine
LASIS: Left anterior superior iliac spine
RPSIS: Right posterior superior iliac spine
LPSIS: Left posterior superior iliac spine

Definition of pelvis coordinate system
O: The origin coincident with the midpoint of the ASISs
Z: The line passing through the ASISs from left to right
Y: The line perpendicular to the plane containing the Z-axis and the line connecting the midpoint of the PSISs and ASISs, pointing superiorly
X: The line perpendicular to the YZ plane, pointing anteriorly

Definition of thigh coordinate system
O: The origin coincident with the right hip centre of rotation
Y: The line connecting the midpoint of femur epicondyle and the right hip joint centre of rotation, pointing superiorly
X: The line perpendicular to the plane containing the Y-axis and the line passing through the femur epicondyle, pointing anteriorly
Z: The line perpendicular the XY plane, pointing to the right

Definition of shank coordinate system
O: The origin coincident with the midpoint of femur epicondyle
Y: The line connecting the midpoint of malleolus and femur epicondyle, pointing superiorly
X: The line perpendicular to the plane containing the Y-axis and the line passing through the medial and lateral malleolus, pointing anteriorly
Z: The line perpendicular the XY plane, pointing to the right

Definition of foot coordinate system
O: The origin coincident with the midpoint of malleolus
Y: The line connecting the proximal 1st metatarsal and the midpoint of malleolus, pointing superiorly
X: The line perpendicular to the plane containing the Y-axis and the line passing through the medial and lateral malleolus, pointing anteriorly
Z: The line perpendicular the XY plane, pointing to the right.

Appendix 2 – Documentation for 3D scanning ethical approval

Imperial College London

Imperial College Research Ethics Committee
Imperial College London
Room 221
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researchethicscommittee@imperial.ac.uk

Professor Anthony Bull
Imperial College London
Department of Bioengineering
London
SW7 2AZ

14 March 2016

Dear Professor Bull,

Study Title: Using a Styku™ 3D scanner technology to obtain subject specific body segment parameters.

ICREC reference: 16IC3224.

The above study was approved by your Head of Department on 08/03/16 and by the Joint Research Compliance Office on the 11/03/16.

Under the Imperial College Research Ethics Committee process, a study that has been reviewed by the Joint Research Compliance Office and Head of Division/Department (or Principal), where no significant ethical issues have been identified in the protocol or ethics application, can be approved without requiring it to go to full committee.

Documents
The documents reviewed were:

- ICREC Application form
- Protocol (v2 10/03/16)
- Participant info sheet (v2 10/03/16)
- Consent form (v2 10/03/16)
- Research poster (v2 10/03/16)
- Participant data form (v1 09/03/16)
- Recruitment email (v1 09/03/16)

Yours sincerely,



Gary Roper,
Head of Regulatory Compliance,
Imperial College London

Version 2, 10th March 2016

Motion Analysis Laboratory
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0747

Investigators: Prof. Anthony Bull

Mr Samuel Smith

Dr Maximilian M Wdowski

Using a Styku™ 3D scanner to obtain subject specific body segment parameters.

Participant Information Sheet

We would like to invite you to join our research study. Before you agree to participate it is important for you to understand why the research is being done and what it will involve. This form gives detailed information about the research study and is yours to keep. Please take the time to read the following information carefully before you decide whether or not you wish to take part. If there is anything that is not clear or if you would like more information please do not hesitate to speak to the investigators.

What is the purpose of the study?

The purpose of this study is to build a database of body segment parameters (mass, centre of mass and inertial properties). This database has the potential to link variables such as segment mass with age, height, weight, any history of bone or joint related pathologies/ disabilities and a variety of factors which affect weight and mobility.

Why have I been chosen?

We would like you to consider participating in this study as you are a healthy person who is aged 18 or above. Please note that if you are unable to speak or read English or suffer from a psychiatric illness that affects your ability to give informed consent then you will be excluded from the study. If you are pregnant then you will also be excluded from this study.

Do I have to take part?

No, your decision to participate is entirely down to you and if you would like to take part, you will be asked to sign a consent form. Even after you have signed this consent form and agreed to join the study, you are free to withdraw from the study at any time. If you decide not to take part, or withdraw from the study, it will not affect any future interactions that you may have with Imperial College. If you choose to withdraw, the information collected up until that point will be retained and used in the study; however, no further data will be collected. Please inform any member of the research team straight away if you no longer wish to participate in the study.

What does participation involve?

Once you have decided to take part in this research, you will be asked to make one, 30 minute visit to the Motion Analysis Laboratory on the South Kensington Campus of Imperial College London, SW7 2AZ, where our research team will discuss the study with you and answer any questions you may have. If you are still happy to

take part, we will ask you to sign the consent form and complete the anonymised participation data form which will ask for information such as age, height, weight, hand dominance, profession, diet/exercise habits and any history of joint/weight related pathologies. We will perform a 3D scan of your body which will involve you wearing skin tight clothing. This will be made available in a variety of sizes or if you prefer, you can bring your own providing they are tight fitting. The scan will take approximately 1 minute. For each scan you will be helped on to, and asked to remain as still as possible on, a rotating platform.

What are the side effects, and are there any risks in taking part?

If you experience any muscle or joint pain and/or discomfort during the session please inform the researcher, and if you need to stop, you can do so at any time. You will need to step onto the platform and the investigator will be on hand to help.

What are the possible benefits of taking part?

There are no clear benefits to you for taking part. However, the scanning equipment we will be using was original designed for fashion purposes and it will be possible to provide you with accurate measurements of your body which we are happy to email to you for your own personal use. The information we get from this research will enable us to create a quick and accurate method of obtaining body segment parameters for subject specific computational models of the musculoskeletal system.

Will my taking part in this study be kept confidential?

Any information you give us will be kept confidential. If the study is published in a book or scientific journal, no individual will be identified in any way. Your personal details will be stored on a computer which belongs to Imperial College London.

What will happen to the results of the research study?

The results of the study will be analysed by the research team and presented at biomechanical and clinical conferences and published in scientific journals. No individual subject will be identified in any report or presentation arising from the research. There are no individual results for this work apart from the data that we are happy to provide for you. You can be provided with a summary report of our findings at the end of the study upon your request.

The data gained from your scan will be stored in a database of body segment parameters alongside the answers you provide with the accompanying participant data form (Appendix 5). As with the individual results, this will remain anonymous.

Will I be paid for taking part in the study?

You will not be paid for your participation in the study.

Who has reviewed the study?

This study was reviewed by the Joint Research Compliance Office (JRCO).

What happens in case of withdrawal?

All the participants will be informed of their right to leave the study at any time without giving any reason. To withdraw from the project the participant has to make it clear to the investigators. This can be done at any point during the project by any of this methods: using the contact email provided in the poster (Appendix 4), contacting one of the investigators using contact information provided in the university webpage

or speaking with one of the researchers. The data collected during the consent period will be stored and no further data of that subject will be recorded.

Contacts for further information:

If you are considering taking part in the study and would like further information please contact Samuel Smith, who may be reached via email at s.smith14@imperial.ac.uk.

Imperial College London

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Version 1, 9th March 2016

Participant data

Using a Styku™ 3D scanner to obtain subject specific body segment parameters.

Before completing this form, please read it through carefully and notify the investigator if you believe you have already completed it for a previous study which took place at any Imperial College London campus.

Subject Identification Number: _____

1. Age: _____
2. Sex: _____
3. Height (m): _____
4. Weight (kg) _____
5. Dominant hand: _____

6. From the options below, please select the best description of your current profession, or if retired, main profession prior to retirement. You may select up to **TWO** options:

Office based/Academia (researcher or lecturer)	☐	Healthcare professional/education (primary or secondary)	☐	Student	☐
Labour/Food industry	☐	Sports/Athletics (please specify)	☐	Performing arts	☐

_____

Armed forces ☐ Retail ☐ Other (please specify) ☐

(please specify) _____

7. Following from the previous question, how many hours in a **typical** working day would you consider your job to be **physically demanding** (heavy lifting, strenuous exercise, etc.)?

   None ☐
   1-3 hours a day ☐
   3-5 hours a day ☐
   More than 6 hours a day ☐

8. Do you have children?

   Yes ☐ No ☐

   a. If 'Yes', how many children do you have? _____

   b. Approximately how long has it been since you had your last child?

      _____ Years _____ Months

9. The following two questions ask about your **current exercise habits outside of work** or the **best description of your exercise habits outside of work in the last 6 months**. Try to consider your answers as a **typical** month/week/day for you. Please also mention any hobbies which could be considered as strenuous movement.

   a. How often do you typically participate in sports/exercise?

      Never
      Once or twice a month ☐
      Once or twice a week ☐
      3 or more times a week ☐

      Please specify which sports/exercise you do:

      _____
      _____
      _____

b. When you participate in any of the above activities do you tend to exercise strenuously, or to exhaustion?

Yes ☐ No ☐

10. The following 5 questions ask about your **current diet** or **the description of your eating habits in the last 6 months**. Try to be as honest as possible for **typical** time periods in your life. Remember all responses will be kept anonymous.

a. How many units of alcohol do you consume in a **typical** week? (A pint of beer/lager/cider is typically 2 (low strength) to 3 (higher strength) units, a standard glass of wine is 2 units and a shot of any spirit is 1 unit).

None ☐
0 – 10 ☐
10 – 20 ☐
20 – 30 ☐
More than 30 ☐

b. How many glasses of water do you drink on a **typical** day?

0 to 3 glasses (0 – 1.5 litres) ☐
3 to 5 glasses (1.5 – 2.5 litres) ☐
5 to 10 glasses (2.5 – 5 litres) ☐

c. How often do you **typically** buy fast food/take away food?

Never ☐
Once or twice a month ☐
Once or twice a week ☐
3 or more times a week ☐

d. Are you currently on a dedicated diet either started by yourself or prescribed by a healthcare professional? This includes dietary habits such as Vegetarianism. Please state the length of time you have been on this diet.

Yes ☐
No ☐

If Yes, Please Specify _____

_____ Years _____ Months

11. Please answer the following question in consideration of your answers to questions 9 and 10. How healthy would you say your life is?

Very Healthy	☐
Healthy	☐
Slightly Healthy	☐
Slightly Unhealthy	☐
Unhealthy	☐
Very Unhealthy	☐

12. Do you currently, or have you ever, suffered from/had any of the following? Tick **ALL** that apply:

	In the past (since recovered) …	Currently suffer from …
Rheumatoid Arthritis	☐	☐
Osteoarthritis	☐	☐
Diabetes (Type 1 or 2)	☐	☐
Back pain	☐	☐
High stress levels	☐	☐
Abnormal tiredness	☐	☐
Chronic heart condition	☐	☐
Obesity	☐	☐
Anorexia	☐	☐
Bulimia	☐	☐
Depression	☐	☐
Chronic Lung condition	☐	☐
Joint replacement surgery	☐	☐
Please specify which joint/s: _____		
Muscle weakness	☐	☐
Please specify region: _____		

13. Please use the space below for anything else you feel will be of interest regarding your general health which may have **affected your weight and/or mobility in the last 6 months**:

Appendix 3 – Documentation for sit-to-stand ethical approval

Imperial College London

Imperial College Research Ethics Committee
Imperial College London
Room 221
Medical School Building
St Marys Campus
London
W2 1PG
Tel: +44 (0)207 594 9484

researchethicscommittee@imperial.ac.uk

Professor Anthony M Bull
Imperial College London
Bioengineering Department
Royal School of Mines
London
SW7 2AZ

22 February 2016

Dear Professor Bull,

Study Title: Observation of kinematics and muscle force distribution patterns during a series of activities of daily living in young, mature and older adults.

ICREC reference: 16IC3181.

The above study was approved by your Head of Department on 09/02/16 and by the Joint Research Compliance Office on the 19/02/16.

Under the Imperial College Research Ethics Committee process, a study that has been reviewed by the Joint Research Compliance Office and Head of Division/Department (or Principal), where no significant ethical issues have been identified in the protocol or ethics application, can be approved without requiring it to go to full committee.

Documents
The documents reviewed were:

- ICREC Application form
- Participant Information Sheet (v3 12/02/16)
- Recruitment email (v2 12/02/16)
- Recruitment poster (v2 12/02/16)
- Consent Form (v3 12/02/16)
- Protocol (v3 12/02/16)
- Sponsorship and Indemnity form
- Edmonton frailty test (v2 12/02/16)
- Scoring system for Edmonton frailty test (v2 12/02/16)
- Participant data form (v2 12/02/16)

Yours sincerely,



Gary Roper,
Head of Regulatory Compliance,
Imperial College London

Version 2, 12th February 2016

Motion Analysis Laboratory
3rd Floor, Royal School of Mines
Department of Bioengineering
Imperial College London
South Kensington Campus
London, SW7 2AZ
Tel: +44 (0)20 7594 0747

Investigators: Prof. Anthony Bull
Mr Samuel Smith
Dr Maximillian Wdowski

Observation of kinematics and muscle force distribution patterns during activities of daily living in young, mature and older adults

Participant Information Sheet

We would like to invite you to join our research study. Before you agree to participate it is important for you to understand why the research is being done and what it will involve. This form gives detailed information about the research study and is yours to keep. Please take the time to read the following information carefully before you decide whether or not you wish to take part. If there is anything that is not clear or if you would like more information please do not hesitate to speak to the investigators.

What is the purpose of the study?

The purpose of this study is to observe the movement patterns used by different age groups during a sit to stand motion and calculate muscle force distribution patterns using computational musculoskeletal (MSK) models. The final aim will be to quantify any observed changes which occur with age and may help to understand how physical frailty affects the MSK system.

Why have I been chosen?

We would like you to consider participating in this study as you are a healthy person who is aged either between 18 and 64 or 65 and 85. You have also been chosen as you have had no surgical interventions on any joint or part of the body which limits mobility and you do not suffer from any systematic inflammatory, connective tissue disorders, any known allergy to adhesives or any other medical disorders that limit exercise. Please note that if you are unable to speak or read English or suffer from a psychiatric illness that affects your ability to give

informed consent then you will be excluded from the study. If you are pregnant then you will also be excluded from this study.

Do I have to take part?

No, your decision to participate is entirely down to you and if you would like to take part, you will be asked to sign a consent form. Even after you have signed this consent form and agreed to join the study, you are free to withdraw from the study at any time. If you decide not to take part, or withdraw from the study, it will not affect any future interactions that you may have with Imperial College. If you choose to withdraw, the information collected up until that point will be retained and used in the study; however, no further data will be collected. Please inform any member of the research team straight away if you no longer wish to participate in the study.

What does participation involve?

Once you have decided to take part in this research, you will be asked to come to the Motion Analysis Laboratory on the South Kensington Campus of Imperial College London, SW7 2AZ, where our research team will discuss the study with you and answer any questions you may have. If you are still happy to take part, we will ask you to sign the consent form. During this visit, you will be asked to complete a short questionnaire which includes a timed up and go test. You will only be required to make one visit to the Laboratory, lasting approximately 90 minutes.

We will ask you to provide us with some information regarding your age, sex, dominant hand, height, profession, sports and diet habits and any history of joint pain or injury (Appendix 5). Once complete, you will be asked to perform a sit to stand action up to 5 times. To analyse the movement of your body we will place adhesive markers on your bare skin at various anatomical locations of your arms, shoulders, torso and legs. You will be encouraged to rest in between repetitions.

What are the side effects, and are there any risks in taking part?

The adhesive used to attach the markers to your skin may cause some redness. If you have any allergies or skin conditions (e.g. eczema, dermatitis, etc.), please inform the researcher prior to the placement of the markers, as you may be unable to take part. If you notice any irritation during or immediately following the study, please inform the researcher and a first aider will be called to assist you. The area will be cleaned with soap and water. If the irritation

persists for more than 24 hours, please contact us and you may be advised to consult your local medical clinic.

If you experience any muscle or joint pain and/or discomfort during the session please inform the researcher, and if you need to rest and/or want to stop, you can do so at any time.

What are the possible benefits of taking part?

There are no clear benefits to you of taking part. However, the information we get from this research will enable us to better understand the effect that ageing has on performance of basic activities of daily living. This information could then be used for tailored rehabilitation to prevent the effects of physical frailty.

Will my taking part in this study be kept confidential?

Any information you give us will be kept confidential. If the study is published in a book or scientific journal, no individual will be identified in any way. Your personal details will be stored on a computer which belongs to Imperial College London.

What will happen to the results of the research study?

The results of the study will be analysed by the research team and presented at biomechanical and clinical conferences and published in scientific journals. No individual subject will be identified in any report or presentation arising from the research. We are unable to provide you with your individual results; however, you can be provided with a summary report of our findings at the end of the study, upon your request.

Will I be paid for taking part in the study?

You will not be paid for your participation in the study.

Who has reviewed the study?

This study was reviewed by the Joint Research Compliance Office (JRCO)

What happens in case of withdrawal?

All the participants will be informed of their right to leave the study at any time without giving any reason. To withdraw from the project the participant has to make it clear to the investigators. This can be done during the research or at the end of it by any of this methods: using the contact email provided in the poster (Appendix 4), contacting one of the investigators using contact information provided in the university webpage or speaking with one of the researchers. The data collected during the consent period will be stored and no further data of that subject will be recorded.

Contacts for further information:

If you are considering taking part in the study and would like further information please contact Samuel Smith, who may be reached via email at s.smith14@imperial.ac.uk.

Imperial College London

Motion Analysis Laboratory
3rd Floor, Royal School of Mines
Department of Bioengineering
Imperial College London
South Kensington Campus
London, SW7 2AZ
Tel: +44 (0)20 7594 0747

Version 2, 12th February 2016

Participant data

Observation of kinematics and muscle force distribution patterns during activities of daily living in young, mature and older adults

Subject Identification Number: _____

Age: _____

Sex: _____

Height (m): _____

Weight (kg) _____

Dominant hand: _____

History of any joint pain or injury:

Appendix 4 – All peak muscle force data for sit-to-stand

Table A. 1 - Peak muscle forces for the upper body whilst using arm rests. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase p<0.0005)

	Mean peak muscle force (% body weight ± SD)					
	Stand			Sit		
Muscle	YO	MA	EL	YO	MA	EL
Trap_clav	7.8 (5.8)	9.5 (8.8)	7.8 (10.2)	9.9 (9.5)	10.9(10.3)	9.4 (10.2)
Trap_scap	37.4(30.6)	29 (25.1)	36.3 (33.6)	42.9(36.5)	32.6(29.8)	43.9 (30.5)
LS	10 (9.2)	9.2 (9.1)	13 (12.8)	16.6 (14)	12.7(13.1)	24.3 (21.5)[b]
Rmin	4.4 (3.6)	3 (3.5)	4.5 (4.5)	7 (5.6)	3.7 (4.3)	8.3 (9.6)[b]
Rmaj	7.3 (6.4)	6.5 (6.9)	8.5 (7.2)	11.6 (9.6)	10 (12.5)	14.5 (12)
SA	88.3(69.9)	87.6 (58.6)	102 (67.5)	124 (77.1)	100 (61.9)	143 (55.4)
Pmin	4.1 (2.9)	5.1 (4.9)	7.4 (6.5)[a]	7 (5.6)	6.7 (7.1)	9.8 (7.7)
LD	5.9 (10.2)	4.6 (7.4)	3.9 (4)	10 (13.9)	10.1(15.9)	5.4 (7.5)
Pmaj_clav	11.7(12.8)	7.7 (9.8)	14.5 (15.6)	14.2(19.2)	9.4 (14.3)	12.8 (16)
Pmaj_thor	53.8(28.6)	40.4 (28)	47.6 (30.1)	38.8(23.1)	31 (22.9)	28.5 (19.6)
DeltA	26.5(15.8)	15.9 (20.8)	19.1 (16.9)	35.7(26.4)	17.7(22.2)[a]	17.1 (16.2)[a]
DeltM	48.8(30.2)	57.3 (45.8)	72 (40.5)	83.5 (40)	73.2(50.4)	98.8 (44.6)
DeltP	20.8(12.8)	24.1 (9.5)	25 (11.5)	25.1(16.3)	25.6(10.1)	28.8 (13.6)
SS	24.6(19.7)	35.8 (31.3)	44.8 (36)[a]	19 (14.1)	36.3(31.5)	51.4 (40.3)[a]
IS	4.4 (7.8)	3.7 (5.8)	2.6 (2.8)	3.1 (5.3)	4.6 (5.5)	2.1 (2.1)
SBS	5.3 (7.9)	10.2 (17.4)	6.4 (12)	4.7 (12.6)	12.4(22.8)	9.8 (24.4)
Tmin	0.2 (0.3)	0.9 (2.4)	0.3 (0.5)	0.2 (0.3)	0.5 (1.2)	0.2 (0.3)
Tmaj	33 (22.6)	49.9 (36.1)	63.2 (35.9)[a]	58.3(31.9)	61.4 (43)	90.1 (45.4)[ab]
CB	1.2 (2)	3.3 (7.1)	1.6 (3)	0.6 (0.9)	1.8 (2.5)	1.2 (1.7)
BIC_short	3.2 (3.9)	2.8 (6.5)	3.1 (3.6)	3.3 (9.7)	3.5 (5.4)	2.9 (3.3)
BIC_long	30.7(25.4)	28.8 (26.9)	46.7 (27)[b]	37.5(24.6)	28.5(24.9)	58.7 (27.1)[aB]
TRI_long	18.5(17.4)	31.9 (25.4)	37.8 (26.6)[a]	27.7(25.7)	35.5(26.2)	52.6 (35.5)[a]
TRI_med	46.3(35.8)	41.6 (30.9)	51 (29)	67.6(41.7)	43.7(34.7)[a]	69.4 (28.2)[b]
TRI_lat	26.1(21.4)	24 (19)	29.9 (17.9)	40.7 (29)	25.5(20.8)	43.6 (21.4)[b]
Bra	5.8 (3.3)	5.8 (6.7)	3.9 (3.7)	5.2 (2.7)	8 (9.7)	4.4 (3.5)
Anc	8.8 (8)	8.7 (7.7)	13.6 (12.5)	13.7 (9.7)	10.7 (9.7)	22.9 (14.4)[ab]
BRD	1.3 (1.1)	1.2 (1.3)	0.9 (0.8)	1.3 (0.8)	1.7 (1.8)	0.9 (0.7)[b]
SUP_hum	0.8 (0.9)	0.8 (1)	1 (1.3)	1.7 (3.5)	0.9 (1)	3.6 (7.8)
PT_hum	0.3 (0.2)	0.3 (0.2)	0.4 (0.4)	0.8 (1.4)	0.4 (0.3)	1.5 (2.6)

Table A. 2 - Peak muscle forces for the lower body during stand. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase p<0.0005). [m] (p<0.05) or [M] (p<0.0005) indicate significant difference between use of arm rests; if placed after muscle name the significant differences were found for all groups

	Mean peak muscle force (% body weight ± SD)					
	Stand			Stand – no arm rests		
Muscle	YO	MA	EL	YO	MA	EL
Add. brev.(prox.)[M]	1.1 (0.7)	1.2 (0.6)	1 (0.5)	2.4 (1.3)	2.7 (1.7)	2.2 (1.5)
Add. brev. (mid.)[M]	1 (0.6)	1.1 (0.5)	0.9 (0.5)	2.2 (1.2)	2.4 (1.5)	2 (1.4)
Add. brev. (dist.)[M]	0.9 (0.6)	1 (0.5)	0.8 (0.4)	2 (1)	2.2 (1.4)	1.8 (1.2)
Add. long.[M]	6.1 (3.8)	6.7 (3.6)	5.5 (2.9)	12.3 (5.3)	14.4 (8.3)	12.3 (7.6)
Add. magn.(dist.)[M]	38.7(8.8)	31.5 (7.6)[a]	26.3 (9)[A]	53.9(10.3)	46 (11)[a]	42.4 (12)[A]
Add. magn.(mid.)[M]	20.7 (4.7)	17.3 (4.3)[a]	14.4(4.6)[A]	29.3 (6.5)	26.1 (5.8)[a]	23.9 (6)[A]
Add.magn.(prox.)[M]	2.6 (0.8)	2.4 (0.8)	1.9 (0.7)	4.3 (1.6)	4.2 (1.3)	3.8 (1.4)
Bic. fem. CL	22.3 (12)	27.3 (16.3)	31.6 (16.5)	25.1 (13.3)	41.4(30.7)[am]	52.1(24.4)[AM]
Bic. fem. CB	3.4 (2.6)	2.8 (2.1)	3.4 (3.5)	3.5 (2.1)	3.1 (2.5)	3.7 (3.6)
Ext. dig. long.[m]	4.4 (5.2)	1.9 (2.5)[a]	1.5 (1.3)[A]	6.7 (7.2)	2.6 (3.6)[a]	1.9 (1.6)[A]
Ext. hal. long.	5.3 (6.2)	2.7 (3.2)	1.8 (1.4)[a]	10 (10.3)[m]	3.5 (4.4)[a,m]	2.1 (2)[A]
Flex. dig. long.	0.7 (0.7)	0.9 (1)	0.9 (0.7)	3.5 (7.5)	1.1 (1.1)[m]	1.3 (0.9)[m]
Flex. hal. long.[m]	4.4 (2.6)	4.9 (2.4)	6.4 (2.7)	17.5 (35.7)	6.4 (3.9)	8.8 (7.6)
Gastrocn. (lat.)	12.6 (7.3)	11.8 (7.1)	12.5 (6.2)	13.5 (7.3)	12.2 (9.5)	13.5 (5.5)
Gastrocn. (med.)	12.9(11.9)	13.6 (8.9)	23.6 (16)[ab]	14 (15.7)	11.5 (6.4)	24.8(26.5)[ab]
Gemellus (inf.)[M]	2 (0.8)	2 (0.8)	1.7 (1)	3.6 (2)	4.1 (4.5)	3.2 (1.3)
Gemellus (sup.)[M]	1.8 (0.5)	1.8 (0.6)	1.6 (0.7)	3 (1.6)	3.3 (3.9)	2.7 (1)
Glut. max. (sup.)[M]	38.7 (12.2)	40.6 (17.1)	32.8 (13.3)	52.3 (15.4)	55.6 (26.5)	50.4 (21.2)
Glut. max. (inf.)[m]	8.5 (3)	7.1 (3.8)	8.3 (6.3)	10.8 (7.2)	9.7 (13.4)	11.2 (14.3)
Glut. med. (ant.)	15 (6.2)	13.5 (6.6)	14.9 (8.3)	15.9 (6.3)	13.8 (7)	15.6 (10)

Glut. med. (post.)	42.3 (8.7)	37.7(10.2)	38.6 (12.3)	43 (10.4)	37.9 (9.1)	39.1 (12.2)
Glut. min. (ant.)	4.5 (2)	4.2 (2.3)	4.5 (2.5)	4.7 (2)	4.2 (2.2)	4.6 (3.2)
Glut. min. (mid.)^{m}	3.3 (1.2)	3.6 (1.4)	3.8 (1.3)	3.8 (2.1)	3.6 (1.4)	4 (1.6)
Glut. min. (post.)^{m}	2.8 (1.2)	3.3 (1.7)	3.8 (2)	3.7 (2.6)	3.9 (2.7)	4 (2.6)
Gracilis^{M}	5.3 (3.6)	3.6 (1.5)^{A}	2.7 (1.2)^{a}	13 (8.5)	7.1 (3)^{A}	5.7 (2.6)^{A}
Iliacus (lat.)	1.7 (1)	3 (1.3)^{a}	3 (1.6)	2.4 (2.2)	3.5 (2.6)^{a}	3 (1.8)
Iliacus (mid.)	4.7 (2.7)	8.2 (3.6)^{a}	8.4 (4.3)^{a}	5.8 (3.9)	9.6 (7.3)^{a}	8.1 (5)^{a}
Iliacus (med.)	2.1 (1.2)	3.7 (1.6)^{a}	3.7 (1.9)^{a}	2.9 (2.4)	4.3 (3.3)^{a}	3.7 (2.3)^{a}
Obt. ext. (inf.)^{M}	3.1 (1.5)	3.1 (1.5)	2.4 (1.2)	5.8 (3)	6.6 (3.7)	5.4 (2.9)
Obt. ext. (sup.)^{M}	21 (11.3)	22.1 (11.1)	17.2 (9.4)	39.1 (20.2)	46.3 (21.1)	37.5 (16.3)
Obturator int.^{M}	26 (10.7)	26.7 (10.7)	21.2 (8.5)	44.7 (20.4)	52.7 (40.5)	41.9 (15.6)
Pectineus^{m}	1.3 (0.8)	1.7 (0.6)	1.8 (0.9)	2.3 (2.3)	2.4 (2.2)	2.1 (1.2)
Peroneus brev.	17 (15.4)	9.1 (10.2)^{a}	5.6 (6.1)^{A}	29.4(26.5)^{m}	10.2 (11.8)^{A}	7.6 (8.4)^{A,m}
Peroneus long.	8.7 (5.9)	9 (6.6)	7.5 (6)	19.1(23.9)^{m}	13.3 (14.4)	7.4 (7.4)^{a}
Peroneus tert.	5.3 (6.9)	3 (4.3)	1.7 (1.9)^{a}	9.9 (10.5)^{m}	3.4 (5.5)^{a}	2.4 (3.1)^{A}
Piriformis^{M}	4.1 (0.8)	3.8 (1)	3.6 (1.1)	4.7 (2.2)	4 (1.2)	4.3 (1.6)
Plantaris^{M}	0.3 (0.3)	0.3 (0.2)	0.4 (0.3)	0.4 (0.3)	0.4 (0.3)	0.5 (0.6)
Popliteus	5.4 (7.6)	2.8 (2.3)	3.3 (3.5)	10.9 (15)^{m}	3.7 (3.9)^{a,m}	4.8 (7)^{a}
Psoas minor^{M}	0.5 (0.3)	0.6 (0.3)	0.5 (0.3)	1 (0.7)	1.3 (1)	1 (0.6)
Psoas major	8.7 (4.9)	15.1 (6.6)^{a}	15.4 (7.9)^{a}	10 (5.7)	17.6 (13.4)^{a}	14.8 (9.1)^{a}
Quadratis fem.^{M}	12.5 (4.2)	12.2 (4.5)	9.8 (3.8)	20.4 (8)	21.5 (6.6)	19.3 (7.4)
Rectus fem.^{M}	84.6 (51.4)	52.1(21.4)^{a}	38.9(21.2)^{A}	140.6 (71)	86.5 (34.1)^{A}	62.3 (26.5)^{A}
Sartorius (prox.)^{M}	4.3 (3.9)	6.3 (4.6)	5.6 (3.4)	6.7 (5.2)	11.3 (7.3)^{a}	11.2 (4.8)^{a}
Sartorius (dist.)^{m}	6.1 (4.5)	9.1 (5.9)	9 (5.6)	9.4 (10.3)	11.7 (13.6)	9.1 (6.7)
Semimembr.	11.4 (5.2)	13.5 (5.9)	15.3 (6.2)^{a}	12.7 (5.7)	18.6(11.6)^{am}	23.8(10.3)^{AM}
Semitend.^{M}	31.7 (23.4)	20.7 (10.5)^{a}	16.1 (5.8)^{A}	62.2 (48.7)	34.5 (21.5)^{a}	28.4 (10.2)^{A}

Soleus (med.)[M]	21.1 (19)	36.2 (27.9)	58.8(43.1)[A]	32.6 (29.7)	50.3 (37.1)	65.6 (42.5)[A]
Soleus (lat.)	52.7 (23.8)	45.4 (28.4)	43 (24.3)	54.3 (20.2)	42.3 (28.2)	56.4 (36.3)[m]
Tensor fasc. l.	6.5 (4.2)	12.1 (7.9)[a]	11.9 (7.4)	7.2 (5.9)	17.1 (25.7)[a]	11.5 (9.3)
Tibialis ant.	22.4 (21.3)	25.4 (19.4)	16.8 (12.7)	42.2(36.5)[M]	40.2(31.3)[m]	17.5 (13.3)[ab]
Tibial post.(med.)[m]	3.3 (3)	6.2 (6.1)	7.3 (6.8)	8.1 (14.1)	10 (12.9)	8.7 (7)
Tibial post. (lat.)[m]	3.3 (3)	6.2 (6.1)	7.3 (6.8)	8.1 (14.1)	10 (12.9)	8.7 (7)
Vastus interm.[M]	75.5 (47.9)	45.6(20.1)[a]	32.1(19.3)[A]	143.5(79.8)	74.8(34.2)[A]	51.2 (31.6)[A]
Vastus lat. (inf.)[M]	8.5 (5.7)	5.8 (2.2)[a]	4.2 (2.1)[A]	17 (12.3)	9.2 (3.6)[a]	7.2 (4.6)[A]
Vastus lat. (sup.)[M]	229.7(119.7)	136.8(58.2)[A]	96.7(53.8)[A]	329.8(158.8)	199.6 (87)[A]	133.1(66.4)[A]
Vastus med.(inf.)[M]	10.9 (8)	9.4 (3.3)	7 (3)[a]	16.3 (11.8)	14.9 (5)	11.7 (5.3)[a]
Vastus med.(mid.)[M]	50.5 (29.9)	36.7(13.1)[a]	26.5(12.6)[A]	88.8 (45.2)	59.1 (20.5)[a]	44.7 (18.8)[A]
Vastus med. (sup.)[M]	42.5 (25.7)	27.2 (10.1)[a]	19.9 (9.6)[A]	81.9 (44.3)	44.9 (17.2)[A]	35.2 (17.4)[A]

Table A. 3 - Peak muscle forces for the lower body during sit. [ab] indicates significant differences to the YO/MA groups respectively (p<0.05, uppercase p<0.0005). m (p<0.05) or M (p<0.0005) indicate significant difference between use of arm rests; if placed after muscle name significant differences were found for all groups

	Mean peak muscle force (% body weight, SD).					
	Sit			Sit – no arm rests		
Muscle	YO	MA	EL	YO	MA	EL
Add. brev. (prox.)[M]	1.1 (0.8)	1.1 (0.6)	0.8 (0.4)	1.8 (1)	2 (1.1)	1.6 (1)
Add. brev. (mid.)[M]	1 (0.6)	1 (0.6)	0.7 (0.4)	1.7 (0.9)	1.8 (0.9)	1.5 (0.9)
Add. brev. (dist.)[M]	1 (0.5)	0.9 (0.5)	0.6 (0.4)	1.6 (0.8)	1.7 (0.8)	1.3 (0.8)

Add. long.[M]	6.4 (3.5)	6.3 (3.6)	4.5 (2.3)	10.3 (4.6)	11.6 (5.7)	9.3 (5.7)
Add. magn. (dist.)[M]	38.7 (8.1)	30.3(6.9)[A]	26.7 (6.6)[A]	51.3 (11)	40 (8.2)[A]	35.9(10.2)[A]
Add. magn.(mid.)[M]	20.8 (4.5)	16.8 (3.9)[A]	14.7(3.4)[A]	27.8 (6.3)	22.6(4.2)[A]	20 (5)[A]
Add. magn.(prox.)[M]	2.5 (0.7)	2.3 (0.8)	1.9 (0.5)[ab]	3.9 (1.2)	3.6 (1)	3 (1)[ab]
Bic. fem. CL	29.5(16.7)	35.2(17.6)	37.9 (13.2)	28.5(14.2)	41.1 (23)[m]	46.3(21.5)[a,m]
Bic. fem. CB	2.4 (2.5)	2.1 (1.8)	2 (2.4)	2.4 (2.1)	1.9 (1.9)	2.2 (2.9)
Ext. dig. long.	5.3 (5.9)	2 (2.3)[A]	1.7 (1.6)[A]	6.8 (7.4)	1.9 (2.8)[A]	1.9 (1.6)[A]
Ext. hal. long.	5.6 (5.3)	2.6 (2.9)[a]	1.9 (2.1)[A]	9.6 (12)[m]	3.1 (3.8)[a]	1.8 (1.9)[A]
Flex. dig. long.	0.7 (0.8)	0.8 (0.9)	0.8 (0.5)	2 (5)	1 (1.5)	1 (0.7)[m]
Flex. hal. long.	3.9 (3.5)	4.6 (3)	5.8 (2.6)	8.5 (18.1)	4.7 (3)	6 (2.4)
Gastrocn. (lat.)	8.5 (7.3)	9.2 (5.9)	8.7 (5.8)	8.8 (7)	7.5 (5.1)	8.6 (5.2)
Gastrocn.(med.)	11 (13.1)	11.9 (9.3)	19.2(14.1)[ab]	10.9 (13)	10.9 (9.1)	21.3(21.4)[ab]
Gemellus(inf.)[M]	2 (0.7)	2.2 (0.9)	1.9 (0.7)	3.4 (1.8)	3 (0.8)	2.5 (0.9)[a]
Gemellus (sup.)	1.9 (0.5)	2 (0.7)	1.9 (0.6)	2.9 (1.4)[m]	2.5 (0.7)[M]	2.3 (0.7)[a,M]
Glut. max.(sup.)[M]	46.3(12.6)	45.6(18.5)	42.1 (16.2)	57.5 (18)	53.5 (19.7)	49.5 (16.4)
Glut. max. (inf.)	8.6 (2.8)	7 (2.8)	7.8 (3.2)	9.5 (5.4)	7 (4)	8.4 (4.6)
Glut. med.(ant.)	14.2 (6.2)	12.8 (6.5)	13.1 (5.3)	14 (5.8)	11.8 (5.7)	13.3 (7)
Glut. med.(post.)[m]	46.1 (9.4)	40.3 (13)	41.6 (11.3)	43.3 (12.9)	36.5 (12.7)	40.7 (12)
Glut. min. (ant.)	4.2 (2)	3.7 (2.1)	3.9 (1.7)	4.2 (1.9)	3.5 (1.8)	3.9 (2.4)
Glut. min.(mid.)	3.4 (1.3)	3.3 (1.5)	3.4 (1.1)	3.5 (2.1)	3 (1.3)	3.4 (1.5)
Glut. min.(post.)	3 (1.1)	3.1 (1.5)	3.3 (1.5)	3.6 (2.7)	2.9 (1.7)	3.1 (1.5)
Gracilis[M]	4.2 (2.6)	3.2 (1.3)	2.3 (0.8)[A]	10.3 (7.5)	6.1 (2.7)[A]	4.4 (1.8)[a]
Iliacus (lat.)	2.2 (2.1)	2.7 (1.3)	2.4 (1.2)	2 (1.6)	2.4 (1.3)	2.2 (1.4)
Iliacus (mid.)[m]	5.8 (5)	7.5 (3.7)	6.7 (3.4)	5 (3.1)	6.5 (3.5)	6 (3.8)
Iliacus (med.)	2.7 (2.6)	3.4 (1.7)	3 (1.5)	2.4 (1.8)	2.9 (1.6)	2.7 (1.7)
Obt. ext. (inf.)[M]	2.8 (1.3)	3.1 (1.7)	2.3 (1.1)	5 (2.5)	5.2 (2.1)	4.1 (2.1)
Obt. ext. (sup.)[M]	18.3 (10.1)	21.6 (12.1)	15.8 (7.3)	33.3 (17.1)	37.6 (15.6)	29.9 (16.2)
Obturator int.[M]	25.2 (8.8)	27.4 (12.1)	23 (8.2)	40.1 (17.4)	41 (12.5)	34.2 (12.8)
Pectineus	1.7 (1.7)	1.7 (0.9)	1.5 (0.8)	1.7 (1.6)	1.6 (0.8)	1.6 (0.9)
Peroneus brev.	16.2 (13.3)	8.2 (9.7)[a]	5.6 (8.2)[a]	25.1(24.9)[m]	8.8 (12.3)[a]	6.1 (6.9)[A]
Peroneus long.[m]	8.2 (8.9)	8.4 (5.9)	6.4 (5.7)	13.7 (15.3)	10.8 (11)	6.2 (6)

Peroneus tert.	6.4 (6.9)	3 (4.1)[a]	1.8 (2.7)[a]	9.4 (11.6)	3 (4.8)[a]	2.1 (2.8)[a]
Piriformis	4.5 (0.8)	4.1 (1.1)	4.3 (1.1)	4.9 (2.3)	3.8 (1.3)	4.3 (1.2)
Plantaris	0.2 (0.3)	0.3 (0.2)	0.3 (0.3)	0.2 (0.3)	0.2 (0.2)	0.3 (0.5)
Popliteus[m]	4.9 (7.9)	2.3 (2.5)	2.3 (2.6)	7.1 (11)	2.6 (3.9)	3.5 (6.3)
Psoas minor[M]	0.5 (0.3)	0.6 (0.4)	0.4 (0.2)	0.9 (0.5)	0.9 (0.5)	0.7 (0.5)
Psoas major[m]	10.4 (8)	13.8 (6.8)	12.4 (6.3)	8.8 (4.9)	12 (6.6)	11.1 (7)
Quadratis fem.[M]	12.1 (3.5)	12.2 (5)	10 (3)	18.5 (6.6)	18.6 (5.5)	15.7 (5.5)
Rectus fem.[M]	72.2 (41.4)	47.5(19.1)[a]	33.3 (11.7)[A]	132.1(71.1)	77.8(32.1)[A]	56.5 (26)[A]
Sartorius(prox.)[M]	4.6 (3.9)	7.2 (5.2)[a]	6.6 (2.9)	6.4 (5.2)	10.5 (6.2)[a]	9.8 (4.8)
Sartorius (dist.)	5.5 (3.4)	7.7 (4.3)	7.1 (4.5)	7.7 (9.2)	6.8 (4.9)	5.9 (5.2)
Semimembr.	16.1 (6.8)	17.4 (6)	18.2 (4.6)	15.6 (5.7)	19.3 (7.8)[m]	21.5 (7.4)[a,m]
Semitend.[M]	25.9 (16.9)	18.7 (7.2)[a]	16.1 (3.9)[a]	55.7 (48.9)	31.7 (20.6)[a]	23.1 (8)[A]
Soleus (med.)[m]	19.7 (18.9)	33.5 (26.3)	58.1(41.3)[Ab]	28.1 (28)	41.1 (35.1)	56.9(38.5)[Ab]
Soleus (lat.)	41.6 (23.8)	41.3 (27.7)	41.8 (26.5)	43 (21.1)	31.3(22.5)[m]	41.8 (24.3)
Tensor fasc. l.[m]	6.6 (4.5)	10.2 (5.7)[a]	9.2 (5.5)	5.9 (4.7)	9.1 (6.5)[a]	7.9 (6.6)
Tibialis ant.	25.3 (17.9)	23.4 (17.6)	15.7 (12.5)	38.9 (31)[m]	34.6(24.9)[m]	13.9(12.7)[Ab]
Tibial post.(med.)	3.2 (3.3)	5.8 (6.1)	7 (7.2)	4 (4.2)	7.8 (9.7)	6.6 (5.2)
Tibial post. (lat.)	3.2 (3.3)	5.8 (6.1)	7 (7.2)	4 (4.2)	7.8 (9.7)	6.6 (5.2)
Vastus interm.[M]	63.8 (35)	38.3(15.8)[A]	27 (12.8)[A]	125.2 (73)	66.7(32.3)[A]	43.5 (27.4)[A]
Vastus lat.(inf.)[M]	7 (3.4)	4.9 (1.6)[a]	3.7 (1.4)[A]	13.9 (10.7)	8 (3)[a]	5.8 (2.8)[A]
Vastus lat.(sup.)[M]	202.1(103.4)	114 (50.2)[A]	83.8 (42.3)[A]	321.5(156.9)	182.8(84.1)[A]	122.7 (65)[A]
Vastus med.(inf.)[M]	9.2 (4.2)	8.5 (2.6)	6.7 (2.2)[a]	15.1 (11.1)	13 (3.8)	10 (3.9)[a]
Vastus med.(mid.)[M]	43.7 (21)	32.3 (10.5)[a]	24.6 (8.7)[A]	78.3 (42.6)	52.6 (17.9)[a]	38 (15.7)[A]
Vastus med. (sup.)[M]	36 (17.4)	24 (8.4)[a]	18.1 (6.3)[A]	69.8 (39.2)	40.2(15.6)[A]	28.9 (13)[A]