BIOMECHANICAL TESTING OF HIP PROTECTORS

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LIST OF ABBREVIATIONS

Abbreviation	Description
AB	Adductor brevis
AI	Anterior inferior
AM	Adductor magnus (page 6)
AM	Anterior middle (page 19)
AS	Anterior superior
BMI	Body mass index
CAD	Computer aided Design
CNC	Computer numerical control
FEA	Final element analysis
GD	Gluteus medius
GN	Gluteus minimus
GT	Greater trochanter
HP	Hip protector
LT	Lesser trochanter
MI	Middle inferior
MM	Center
MS	Middle superior
OE	Obturator externus
OI	Obturator internus
OA	Osteoarthritis
PI	Posterior inferior
PM	Posterior middle
PS	Posterior superior
SD	Standard deviation
STEP	Standardized graphic exchange
3D	Three dimensional
TI	Tendon of iliopsoas
VL	Vastus lateralis

UJIAN BIOMEKANIKAL PELINDUNG PINGGUL

ABSTRAK

Kejadian retakan osteoporotik merupakan trend yang semakin meningkat setiap tahun di seluruh dunia. Insiden utama fraktur osteoporosis adalah keretakan tulang pinggul. Kebiasaannya, pad pelindung pinggul yang bersifat keras atau lembut dimasukkan ke dalam seluar atau seluar dalam. Ia direka untuk melindungi kawasan pinggul bagi mengelakkan keretakkan pinggul apabila jatuh pada arah sisi. Kajian ini bertujuan untuk menguji keberkesanan pelindung pinggul dalam mencegah berlakunya keretakkan pinggul. Rig experimen telah direka bentuk untuk mensimulasikan parameter yang sama sepertimana situasi jatuh yang sebenar dari ketinggian yang boleh menyebabkan keretakan pinggul. Dua jenis pelindung pinggul telah diuji iaitu pelindung pinggul lembut dan pelindung pinggul keras. Daya ambang untuk menyebabkan keretakkan tulang pinggul dikenalpasti bernilai 2.5 kN dan pelindung pinggul digunakan untuk mengurangkan daya impak apabila jatuh pada arah sisi, seterusnya menghalang pinggul dari patah. Berdasarkan keputusan eksperimen, semua pelindung pinggul mampu mengurangkan daya yang diterima oleh tulang pinggul di bawah daya ambang yang boleh menyebabkan berlakunya keretakkan pinggul. Pelindung pinggul lembut didapati mengurangkan lebih banyak daya berbanding pelindung pinggul keras. Pelindung pinggul yang berada di pasaran (HipSaver) dapat mengurangkan sebanyak 7.36% lebih daya impak dan pelindung pinggul lembut (LD45 busa polyethylene) mengurangkan sebanyak 6.31% lebih daya impak berbanding pelindung pinggul keras. Pelindung pinggul lembut juga lebih ergonomik disebabkan saiz yang lebih nipis, ringan dan fleksibel. Sebagai kesimpulan, pelindung pinggul adalah alternatif luaran yang berkesan yang boleh digunakan untuk mengurangkan kejadian patah pinggul dan menambah baik gaya hidup pesakit osteoporosis.

BIOMECHANICAL TESTING OF HIP PROTECTORS

ABSTRACT

The occurrence of osteoporotic fracture is an increasing trend yearly around the world. The main incidence of osteoporosis fracture was hip fracture. A hip protector is a specialized pad either hard or soft usually inserted into pants or the underwear. It is designed to protect the hip region to prevent hip fractures following a fall. The purpose of this study is to test the efficacy of hip protectors in preventing the occurrence of a hip fracture. An experimental rig was set up to replicate the parameters similar to an actual sideway fall from a standing height that is able to cause a hip fracture. Two types of hip protectors was tested namely the soft hip protector and the hard hip protector. The threshold force to cause a hip fracture is identified to be around 2.5 kN and the hip protectors is used to attenuate the impact force from the sideway fall preventing the hip from fracturing. Based on the results, all hip protectors was capable to attenuate the force and reduce the force received by the femur below the threshold for hip fracture. Soft hip protectors are found to attenuate more force compared to the hard hip protectors. The market hip protector (HipSaver) can attenuate 7.36% more impact force and the soft (LD45 polyethylene foam) hip protector attenuates 6.31% more impact force compared to the hard hip protector. The soft hip protectors are also more ergonomic due to its thin, lightweight and flexibility. In conclusion, hip protectors are an effective external alternative that can be used to reduce the occurrence of hip fractures and improve the lifestyle of osteoporotic patients.

CHAPTER ONE
INTRODUCTION

1.1 Background

Hip fractures represent a severe health problem for the elderly. Hip fracture can be defined as a bone fracture that occurs at the proximal (upper) site of the femur, at the outer area where femoral head (ball) meets the acetabulum (socket) within the pelvis. Hip fracture can be generally classified into three major types based on anatomical sites: femoral neck, intertrochanteric and sub trochanteric fractures [1, 2]. Studies suggest that the impact force applied to the greater trochanter area during a fall causes the majority of hip fractures. In many countries, large increases in hip fracture incidence are expected due to increasing life expectancy and ageing population [3].

Epidemiological studies have estimated that there would be an exponential increase in the incidence of osteoporotic fractures in Asia, so that by 2050, 50% of all hip fractures would occur in this region [4]. In Malaysia, in 1997, the incidence of hip fracture among individuals above 50 years of age was 90 per 100,000 population [5]. The incidence increased with age; in the 50–54 year olds, the incidence was 10 per 100,000, rising to 510 per 100,000 in those over 75 years old [5]. Approximately 20% of older adults hospitalized for a hip fracture die within a year and about 50% will suffer a major decline in independence [6].

The force to cause a hip fracture varies from one individual to another. Many factors such as age, sex and body mass index can affect the possibility of a hip fracture from occurring. Elderly people are generally more prone to hip fracture compared to younger people due to the reduced upper body strength, coordination and speed. The slow protective response of the body to cushion the impact or break the fall by extending the arms will result in higher force being applied to the hip [7]. The average force taken that will cause a femur fracture is 2.5 kN [7-9]

Hip protectors represent a promising solution in preventing fall-related hip fractures. There are two types of hip protectors, namely hard shell hip protectors and soft shell hip protectors. Hip protectors with energy-absorbing or energy-shunting properties have been designed for active prevention of hip fractures. However, their clinical effectiveness is still under debate [10, 11] and tends to depend on the severity of the fall and the compliance of the user [12, 13].Clinical trials have yielded conflicting results due, in part, to lack of agreement on techniques for measuring and optimizing the biomechanical performance of hip protectors as a prerequisite to clinical trials [6].

The purpose of this project is to create an experimental setup which adhere closely to the general consensus from the International Hip Protector Research Group for biomechanical testing [6]. The experimental setup should be able to represent the conditions of a sideway fall on the greater trochanter which results in a hip fracture. As for the hip protector, most of the hip protectors available in the market comes in standard sizes. However based on a study to design a hip protector for the Korean elderly, it is reported that the Korean elderly has smaller frame in parts of the hip joint compared to Westerners [14]. There is a possibility that the efficacy of the hip protectors to prevent hip fractures being reduced due to improper fitting contributed by the difference in hip size and hip surface geometry [15]. In this research paper, a hip protector will be designed according to the 3D mapping of the hip of the user. Our hypotheses is that with a better fit, the hip protector will remain in position during falls and better attenuate the impact force from the fall.

1.2 Problem Statement

There is no clear standard for the biomechanical testing for hip protectors. It is difficult to predict the impact force applied to the greater trochanter during a sideway fall. The force applied may differ due to the muscle response of an individual and the surface at which the individual falls on. Factors such as the effective mass, hardness of the impact surface and the natural damping effects of the body can vary the impact force. This project is to design an experimental rig which is able to represent a sideway fall from standing height on the human hip.

Apart from that, despite studies that have been conducted to prove the effectiveness of hip protectors, there is still debate on the efficacy of hip protectors. Examples of the issues related to the debate over the efficacy of hip protectors are the different testing methods, the fit of the hip protector to the user, user compliance and comfort. These concerns will be validated thru experimental testing using the designed experimental model.

1.3 Objective

The objective of this research is:

1) To design and fabricate an anatomical hip model which has similar geometrical and physical properties with an actual human hip.
2) To develop a hip protector that specifically fit the hip geometry of the user and compare it with the performance of other hip protectors to attenuate the impact force of a sideway fall below the fracture threshold of the femur bone of 2.5 kN.

1.4 Scope of Work

This project involves fabrication, experimentation and analysis of the results. The drop impact tester fitted with an impact plate is crucial in this experiment, to deliver an impact force similar to an impact force of a sideway fall on the hip. By manipulating the slotted mass and the drop height of the mass, the conditions for an impact force of a sideway fall from standing height on the hip can be achieved. The available drop impact tester from the School of Materials and Mineral Resources of Universiti Sains Malaysia (USM) does not have an impact plate and the slotted mass available are limited. Therefore, the impact plate with a specific diameter and the slotted mass was fabricated.

Apart from that, another important part of this research is the anatomical model of the human hip. The anatomical hip model must exhibit properties similar to an actual human hip in order to perform the biomechanical testing of the hip protector. The fabrication of the anatomical hip model can be divided into two parts which is the bone (profile of greater trochanter) and the human tissue. The profile of the greater trochanter was made of aluminium and the human tissue was made using polyethylene foam of different densities which has similar shore hardness to human tissue. The anatomical model was placed under the drop impact tester to be tested with and without the hip protector. The anatomical hip model was able to receive an impact force similar to a sideway fall with slight damage.

Two types of hip protectors were fabricated, a hard shell hip protector which shuns the impact force and a soft hip protector to absorb the impact force. The hard hip protector was designed according to the surface geometry of the hip to ensure that the hip protector will fit according to the user's hips. The soft hip protector was made with a soft material with high impact absorbing properties. The width and height of the hip protector was designed using the dimensions of existing market hip protectors taken from previous published papers. Performance testing of the designed hip protectors were conducted by placing it on the anatomical hip model under the drop impact tester to record the amount of force it attenuates.

1.5 Thesis Outline

This thesis is divided into five chapters. Chapter One provides an overview of the severity of hip fractures among the elderly and the proposed solution which is using a hip protector. The objective of this project was clearly stated and the scope of work which this project will be covering was briefly explained. Chapter Two determines the best method or the methods best suited to the availability of the resources in the university such as machines and the raw materials. Chapter Three highlights the important stages in completing the project such as the fabrication process, the experimental setup and the impact testing. The results from the impact testing was then analyzed and discussed in Chapter Four. Finally, the overall findings, results of the project and future recommendations are reinforced in Chapter Five.

CHAPTER TWO
LITERATURE REVIEW

2.1 Overview

In this section, the anatomy of the hip is described and the critical regions that increase the risk of hip fractures are identified. The amount of force to cause a hip fracture and the biomechanical testing of hip protectors was also reviewed to evaluate the suitability of the experimental rig used to test the hip protectors. The anatomical hip model is evaluated based on its degree of representation of an actual human hip and its testing method. It is crucial that the anatomical hip model closely represents the human hip as the credibility of the hip protectors depends on the experimental setup.

2.2 Hip Anatomy

The hip joint is a “ball-and-socket” type joint that is composed of bone structures (femoral head and acetabulum), fibro cartilaginous structures (acetabular labrum), cartilage layers covering the hip joint, capsular ligamentous structures, synovial joint, muscles and tendons, synovial bursae, neurovascular structures [16]. The hip joint provides both stability and multiaxial mobility. It is also involved in transferring the weight of the body. It is a common site of osteoarthritis (OA) and its incidence is rising as the population ages [17]. Figure 2.1 shows the main features of the femur bone and Figure 2.2 shows the cross-sectional coronal view of the anatomy of the hip joint.

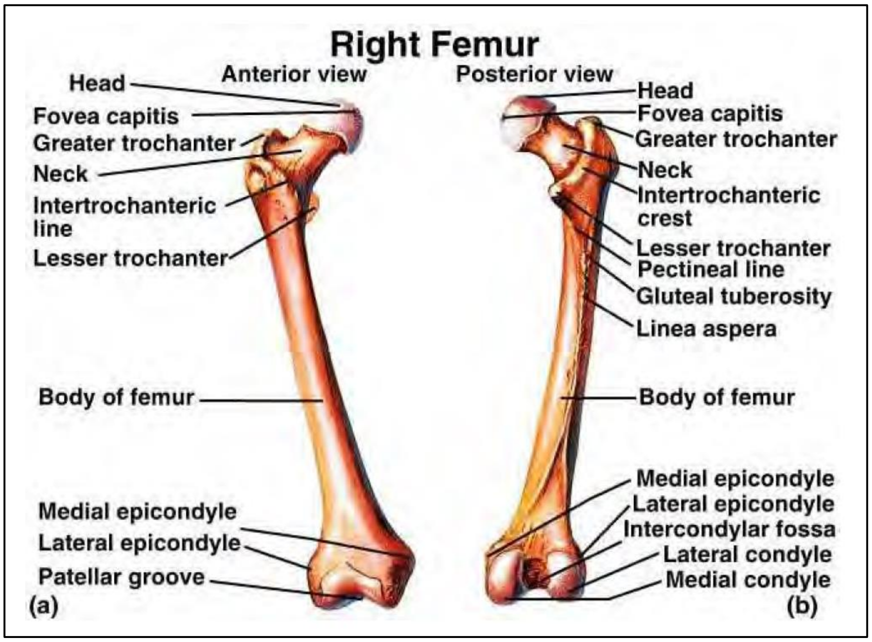

Figure 2. 1: Anatomy of the right femur (a) anterior and (b) posterior view (available online at www.ma.psu.edu).

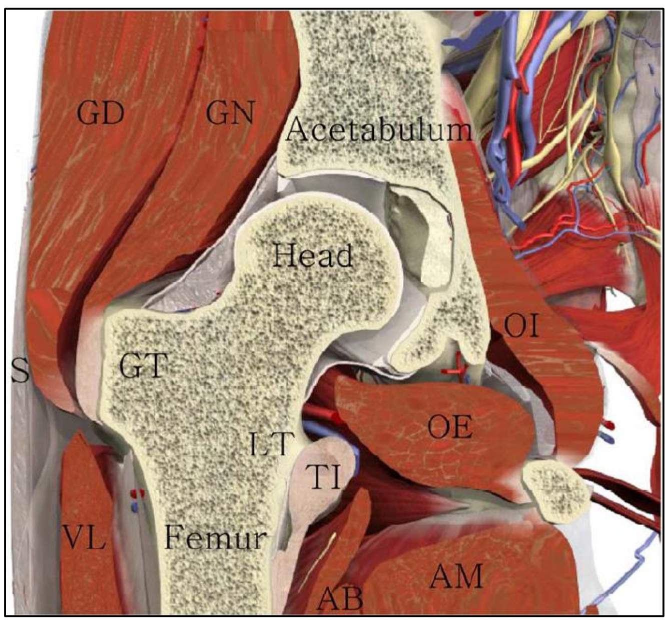

Figure 2. 2: Cross-sectional coronal view of the anatomy of the hip joint (Accessed from www.primalpictures.com)

Various muscles span the hip joint and contribute to the hip joint movement. GD = gluteus medius, GN = gluteus minimus, AM = adductor magnus, AB = adductor brevis, VL = vastus lateralis, OE = obturator externus, OI = obturator internus, TI = tendon of iliopsoas, GT = greater trochanter, LT = lesser trochanter, S = skin and fat.

2.3 Hip Fracture Risk

Based on the report on osteoporotic fracture in Malaysia from the year 1996–1997, those aged over 50 years showed the main incidence of osteoporotic fracture was hip fracture [5]. In the year 2009, it was reported by the National Orthopedic Registry of Malaysia that there were around 510 hip fracture cases being reported. There were a total of 345 female patients and 165 male patients. The demographic showed that around 8.6% patients with osteoporosis were aged between 50 to 59 years old, 20% were aged between 60 to 69 years old, 41.4% were aged between 70-79 years old, 23.9% were aged between 80-89 years old and the remaining 6.1% were aged over 90 years old (Figure 2.3). The age group with the highest osteoporosis cases are those aged between 70 -79. Among the three main races in Malaysia, the Chinese ethnicity has the highest incidence of hip fractures compared to the Malays and Indians, accounting for 44.8% of hip fractures in women [18].

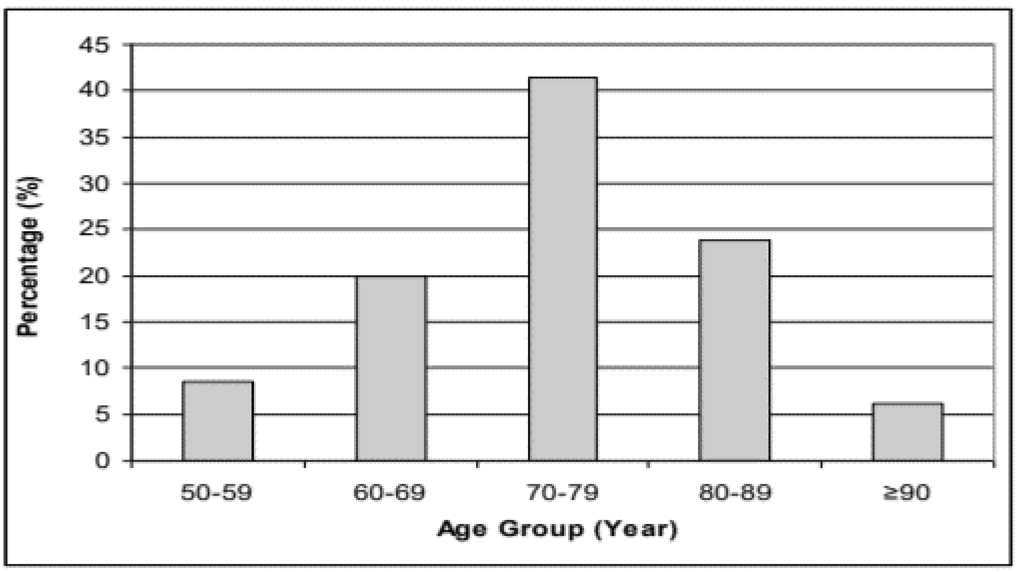

Figure 2. 3: Distribution of hip fracture patients by age group [18]

In a more recent survey conducted at a tertiary private hospital in Malaysia over a period of 5 years (2010 – 2014). There was 258 presumed osteoporotic fractures. Out of 258, 193 were female and 65 male. The median age was 79.0 years (interquartile range [IQR], 12.0 years). Once again among the three main races, the Chinese ethnicity reported the highest number of hip fractures, followed by Indians and Malays. There were 200 Chinese (77.5%), 31 Indians (12.0%), 20 Malays (7.8%), and 7 other races (2.7%). There were 35 patients (12.6%) who were noted to have had a previous low-trauma fracture, of whom 4 received medication [19].

Most hip fractures are due to a direct impact on the trochanteric area of the hip due to sideways fall from standing height [20, 21]. The impact force increases directly with the body weight and falling height of the body and its effect varies with the degree of padding on the greater trochanter by soft tissue and clothing [22-25]. There are three major types of hip fracture classifications based on anatomical sites, namely the femoral neck, intertrochanteric and subtrochanteric fractures [1, 2] as shown in Figure 2.4.

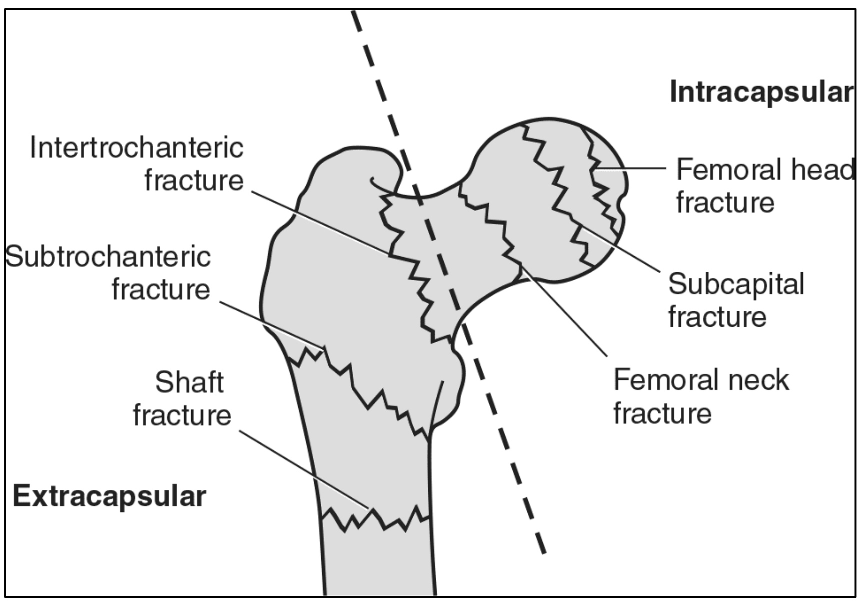

Figure 2. 4: Various types of fractures (available online at fadavispt.mhmedical.com)

In Figure 2.5, the greater trochanter which is highlighted in a red circle has to deal with a high amount of strain after a sideway fall on the hip. The only part that has a higher strain level is the pubic symphysis. This pubic symphysis is a ligament tissue, which is the fibrous connective tissue that connects bones together. In the test, the pubic symphysis is assigned the same mechanical properties as soft tissues [26]. Therefore, the most likely place for a fracture to occur is at the trochanter.

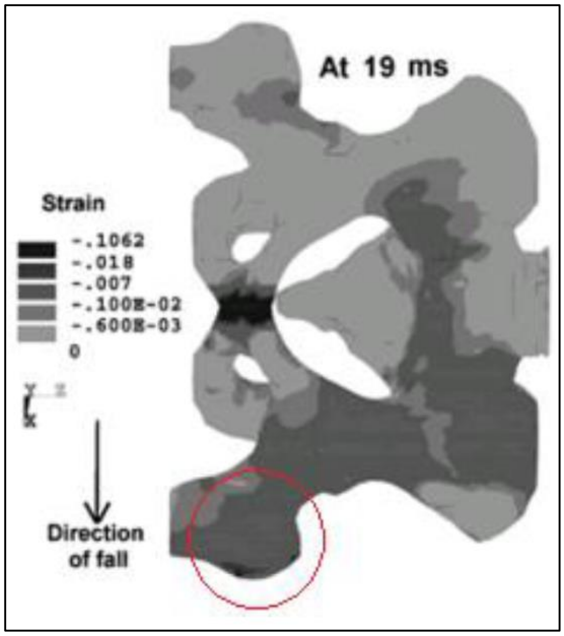

Figure 2. 5: FEA of a sideways fall on the hip [22]

2.4 Determination of Impact Force

In order to generate the impact energy of a sideway fall on the hip, all test systems use a falling mass. There are two common methods in which mass falls vertically by means of a drop impact tower (Figure 2.6) [3, 27] or in a curved path by a pendulum (Figure 2.7) [7]. Precaution is required in the case of a drop impact tower to avoid binding between the falling mass and the guides during impact. On the other hand, the case of the pendulum, the effective mass and any compliance in the pendulum arm must be factored in when calculating the total effective mass and stiffness. Sensors are required to accurately measure the impact velocity of the mass and match this to the desired fall velocity for both methods [6].

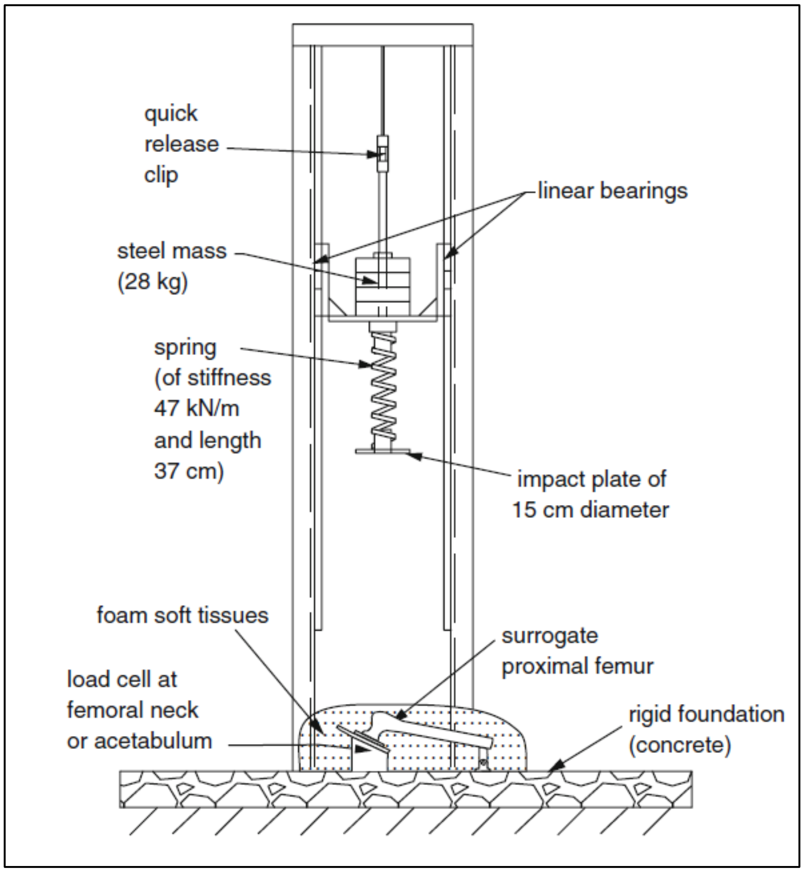

Figure 2. 6: Drop impact tower [6]

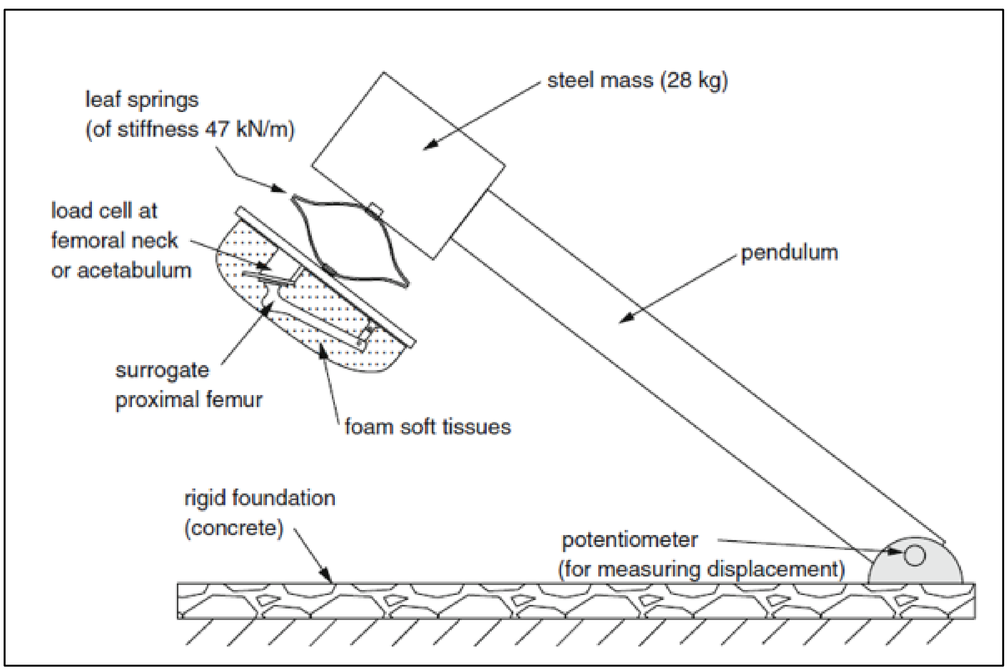

Figure 2. 7: Pendulum Impact [6]

The amount of impact force varies as it is dependent on many factors such as the factors of the human body itself and the impact surface in which the fall occurs. Factors of the human body such as age, weight, height, gender, the amount of soft tissue covering the trochanter and body mass index (BMI) are all related to the fracture force limit [28].

An obese person may have a greater amount of body fat to dissipate the input energy. However, there is a possibility that the increase in the fraction of body mass participating in the impact during the fall may surpass the bone fracture threshold. The same goes to those who are taller. They have greater effective mass and are subjected to higher input energies during the fall [29, 30]. The height dependent change from potential energy to kinetic energy is related by standard laws of one dimensional motion expressed by Equation 2.1.

$$v = \sqrt{2\mathrm{gh}} \tag{2.1}$$

Where v = Velocity (ms^{-1})
g = Gravitational acceleration (ms^{-2})
h = Drop height (m)

As for the impact surface, the flooring on which he or she falls can be hard or soft and the amount of clothing the person wears can have a positive effect. Since there are many combinations of factors, it is difficult to obtain a specific number for the fracture force threshold [28].

Robinovitch, et. al., 1995, a surrogate human pelvis was used to conduct simulated fall impact experiments on trochanteric soft tissues harvested from the cadavers of nine elderly individuals. For each impact, the total applied energy was 140 J. Peak forces ranged from 4,050 to 6,420 N, and tissue energy absorption ranged from 8.4 to 81.6 J [25]. Melo et al., 2010, using a pendulum impact machine, subjected to a high impact energy of 120 J similar to other studies [7-9].

Other experimental studies such as voluntary natural fall from standing height by young adults, pelvis-release experiments to measure the damping properties of hip soft tissues and predict the impact force (Figure 2.8) [31], and fall from a kneeling position to measure the impact force in a low-severity fall have been conducted (Figure 2.9) [32]. Dynamic models have also been developed to determine the fall-induced impact force [33]. A summary of the impact force from various studies are shown in Table 2.1 [33].

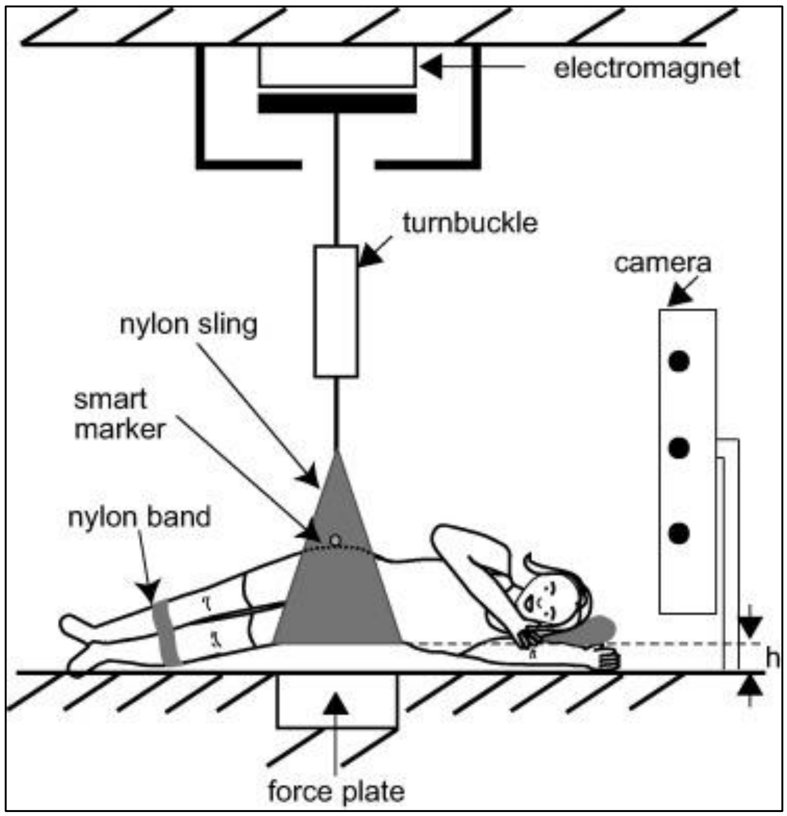

Figure 2. 8: Schematic of equipment and participant positioning during the lateral pelvis release experiments [31].

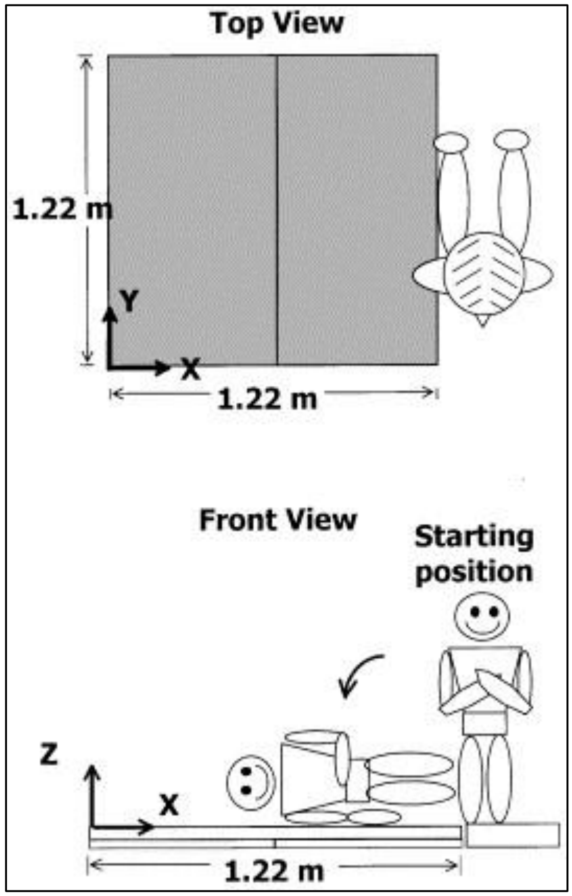

Figure 2. 9: Schematic of participant falling from a kneeling position [32]

Table 2. 1: Results from studies reporting the range of impact velocity and force it falls on the hip [33]

Study	Mean (SD or range) age in years, sample size			Method	Range of femoral force (N)	Range of impact velocity (m/s)
	Men	Women	Mixed			
Robinovitch et al. [13]	28 (5), $n = 7$	26 (6), $n = 7$	27 (6), $n = 14$	Dynamic and impact model	Fall height 0.7 m: 5600 – 8600	1.40 – 6.07
Kroonenberg et al. [14]	–	–	–	Dynamic and impact model/ experiments	3720 – 9990 dynamic model 2900 – 4260 experiments	3.35 – 4.34 dynamic model 2.47 – 2.93 Experiments
Robinovitch et al. [15]	$n = 3$	$n = 6$	77 (10), $n = 9$	Impact pendulum experiments	4050 – 6420	–
Kroonenberg et al. [16]	23.7 (3.7), $n = 6$			Natural sideways fall from standing height	–	2.14 – 4.79
Hayes et al. [17]	$n = 6$	$n = 6$	20 – 35, $n = 12$	Fall experiments from a height of 0.7 m	Men: 6100 – 12 , 100 Women: 5050 – 6370	–
Robinovitch et al. [18]	$n = 5$	$n = 5$	25 (4), $n = 10$	Dynamic models and pelvis-release experiments	1145 – 5288	–
Robinovitch et al. [19]	–	–	–	Impact pendulum experiments	1700 – 5600	1.16 – 2.58
Sanndler and Robinovitch [20]	–	–	–	Pendulum dynamic model	–	1.34–4.14
Robinovitch et al. [21]	–	23 (5), $n = 22$	–	Sideways fall experiments	160 – 387 J backward rotation 6 – 291 J forward rotation	0.58 – 3.71
Robinovitch et al. [22]	–	24 (5), $n = 23$	–	Fall experiments	–	3.3 ± 0.3
Feldman and Robinovitch [23]	$n = 13$	$n = 31$	21 (2), $n = 44$	Sideways fall experiments	–	3.01 (0.83)
Laing and Robinovitch [24]		23.1 (2.4), $n = 14$		Pelvis-release experiments	1004 – 3434	
Levine et al. [25]	$n = 14$	$n = 14$	22.4 (2.7), $n = 28$	Pelvis-release experiments	1415 ± 235	–
Choi et al. [26]	–	–	–	Pendulum experiments for simulating moderate falls	3.465 (143)	2.0
Nasiri and Luo [27]	39.8 (22.1) , $n = 50$	53.7 (19.9) , $n = 80$	48.3 (21.8) , $n = 130$	Subject-specific dynamic and impact model	1883 – 5317	3.39 – 4.31
Average across the whole studies	37.0 (10.3)	40.1 (7.7)	39.1 (8.1)	–	5200	3.0

2.5 Femoral Strength

To appropriately evaluate the efficacy of the hip protector, it must reduce the peak compressive force at the proximal femur during a simulated fall below the value that is expected to cause a fracture. The amount of force required to fracture the femur varies depending on the direction of the force acting on the femur. However, in this study, only the force normal to the greater trochanter is considered.

Courtney et al. 1995, reported that the mean fracture force was 52% lower for older than young adults (3,440 N versus 7,200 N). The femora from the cadavera of eight older individuals with a mean age of seventy-four years and nine younger individuals with a mean age of thirty-three years was mechanically tested using fall loading configuration (Figure 2.10) which represents a fall on the greater trochanter.

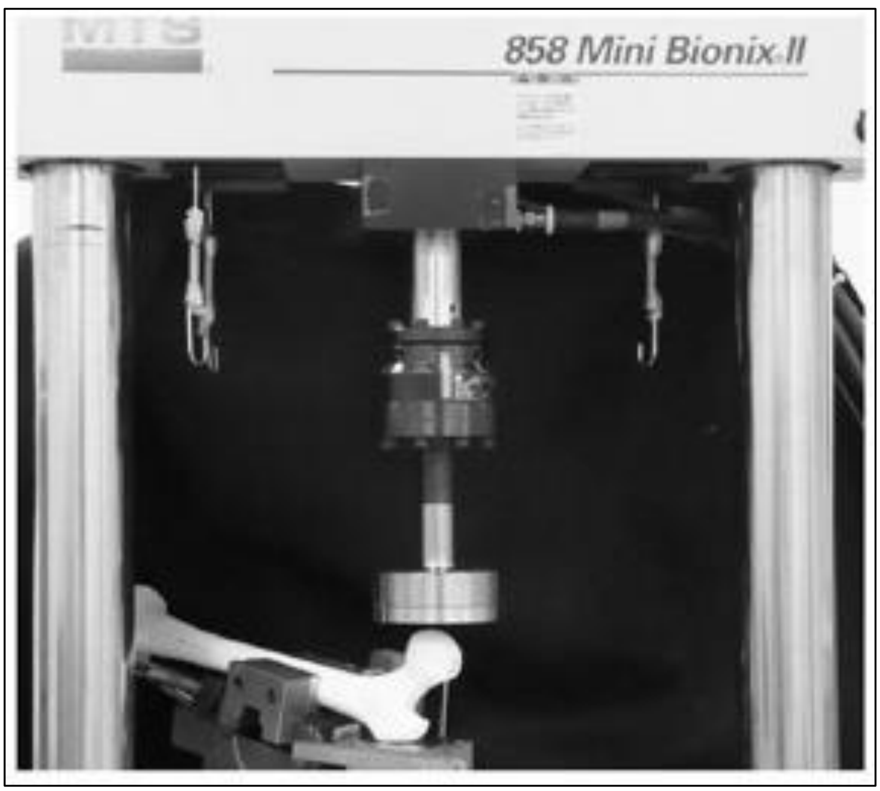

Figure 2. 10: Biomechanical test of the femur neck using the simulated fall configuration [34]

A study by Robinovitch et al. 1995, found that in order to produce a hip protector pad that can attenuate the impact force below the femur fracture threshold of 2.5 kN was to combine energy shunting properties and energy absorbing properties. Nabhani et al. 2002, also compared the performance of a new combination hip protector design with existing designs using a test rig similar to a drop impact tester to simulate the conditions of a sideway fall with a femur fracture threshold value of 2.5 kN. Melo et al. 2010, used an impact pendulum with a set value of 2.5 kN as the threshold of fracture of the femur bone to evaluate the performance of hip protectors made of composite materials. The fracture threshold of 2.5 kN was set for comparison purposes. The hip protectors studied in these works were considered capable of reducing the impact load to the safe range when the impact force is reduced below this threshold value.

Robinovitch et al. 2009, analyzed the results from 14 studies (Table 2.2) that have reported the fracture force of the elderly cadaveric proximal femur tested in a fall loading configuration [6]. The results indicate that age and gender have a significant effect on the femoral fracture strength. The median femoral strength averaged across all studies for studies in which male and female data were combined was 3,472 N (range, 2,110 to 4,354 N), and the median standard deviation was 1,534 N (range, 695 to 1,886 N). Apart from that, studies that reported age-specific values [20], the mean femoral strength was approximately 50% lower for specimens from older than from younger adults (3,770 N for specimens of mean age 74 years (SD=7 years) versus 7,550 N for specimens of mean age 33 years (SD=13 years)). The median femoral strength of specimens of older adults (median age = 82 years for female and 78 years for male), was approximately 30% lower for female than male specimens (2,966 versus 4,220 N).

Table 2. 2: Results from studies reporting the strengths of the cadaveric proximal femur from older adults in a sideway fall loading configuration [6].

		Mean(SD) fracture force (N)			Mean(SD or range) age in years, sample size		
Study	Condition	Women	Men	Mixed	Women	Men	Mixed
Lotz and Hayes, 1990[c]				2,110(1,060)			69(9); n=24
Courtney et al. 1994[c]	Deformation rate=100 mm/s			4,100(1,600)			74(7); n=8
	Deformation rate=2 mm/s			3,440(13,30)			74(7); n=8
Bouxsein et al. 1995[c]				3,680(1,540)			76(59-96)[b]; n=16
Pinilla et al. 1996[c]	0° Load angle			4050(900)			79(11); n=11
	15° Load angle			3,820(910)			81(7); n=11
	30° Load angle			3,060(890)			74(11); n=11
Cheng et al. 1997,1998[d]		3140(1240)	4630(1550)	3,980(1,600)	71(15); n=28	67(15); n=36	69(15); n=64
Bouxsein et al. 1999[c]		1997(1127)	3593(1614)	2,636(1,534)	82(13); n=16	78(10); n=10	81(12); n=26
Keyak et al. 2000[e]				2,400[a]			70(52-92)[a]; n=17
Lochmuller et al. 2002[d]		3,070(1060)	4,230(1530)		82(9); n=63	76(11); n=42	
Eckstein et al. 2004[d]				3,925(1,650)			79(11); n=54
Heini et al. 2004[c]				2,499(6,95)			76(7); n=20
Manske et al. 2006[c]				4,354(1,886)			69(16); n=23
Pulkkinen et al. 2006[d]		2,821[a]	4,209[a]	3,472[a]	82; n=77	79; n=63	81; n=140
Bouxsein et al. 2007[c]				3,353(1,809)			81(11); n=49
Pulkkinen et al. 2008[d]	Cervical fx	2,879(1,117)	4,079(1,165)		82(11); n=34	78(11); n=28	
	Trochanteric fx	3,053(976)	5,506(1374)				
Across study average		2,827	4,375	3,392	80	76	76

[a] SD not provided

[b] Range (not SD) reported

[c] Specimens were stored fresh-frozen

[d] Specimens were embalmed in alcohol/formalin

[e] Specimens were stored frozen, but the authors did not specify fresh versus embalmed.

2.6 Design and Geometry of Hip Model

i) Femur

In this section, the anatomical hip model can be separated into two parts. The first part is the femur, specifically the profile of the greater trochanter and the second part is the human tissue surrounding the bone. In a previous study, the femur model was made of steel and the artificial pelvis was made of aluminum. Steel was used instead of a bone like material because of its robustness. Aluminum, which was chosen to minimize the weight of the apparatus. Metal structures are also used to represent the mechanical hip because it simulates the force transmission, but not the deformation behaviour of human hip bones [3, 6]. Apart from steel and aluminum, second generation composite bone can be used, in which E-glass/ Epoxy Composite simulates cortical bone, and Rigid Polyurethane Bone simulates cancellous bone (Sawbones Europe AB, Sweden) (Figure 2.11) [27].

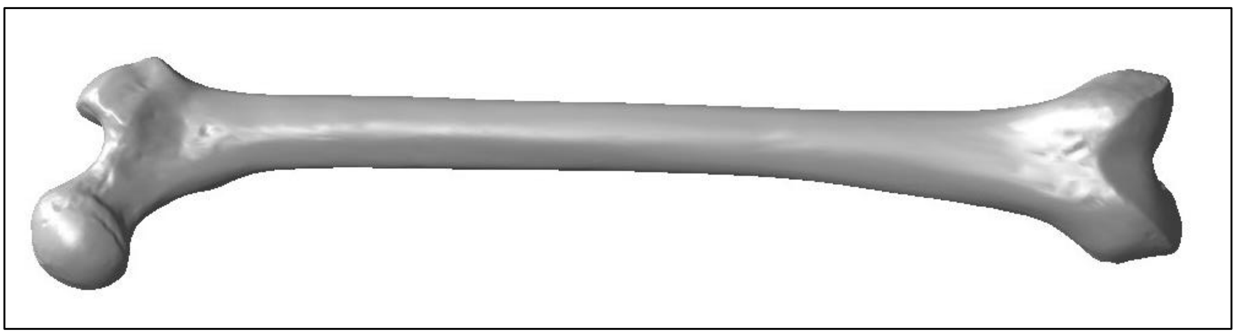

Figure 2. 11: External anatomy of femur with condylar notch (available online from: www.sawbones.com)

ii) Human Tissue

As for the human tissue, materials similar to human thigh flesh required identifying several characteristics of human flesh such as density, hardness, and impact response. The density, hardness, and peak impact load for human thigh flesh were found to be 950 kg/m3, 13.8 (Shore A scale), and 623 N, respectively. Of the candidate materials tested, a custom variation of an existing material known as Sorbothane which has a combination of properties which most closely matched those of human flesh [35]. Another material with a density of 1230 kg/m3 (Wacker Elastosil M

4511), which showed a material behaviour similar to relatively stiff human hips at quasi-static conditions can also be used as human flesh surrogate [3]. Besides that, in a similar study, a closed-cell polyethylene foams, Plastazote HD80 of density 80 kg/m^3 and LD45 of density 45 kg/m^3, was used directly over the proximal femur (Figure 2.12) and closed-cell copolymer foam, Evazote EV50 with a density of 50 kg/m^3, over the regions anterior, posterior, and superior to the femur. These materials were glued together to form a single 21.6 × 24.5 × 8.0 cm^3 block. A 1.2 cm thick layer of open-cell ester foam (SCH180-60E1 of density 29 kg/m^3) was secured over the entire outer surface of the pelvis, along with a 1.6 mm layer of gum rubber to simulate skin [13].

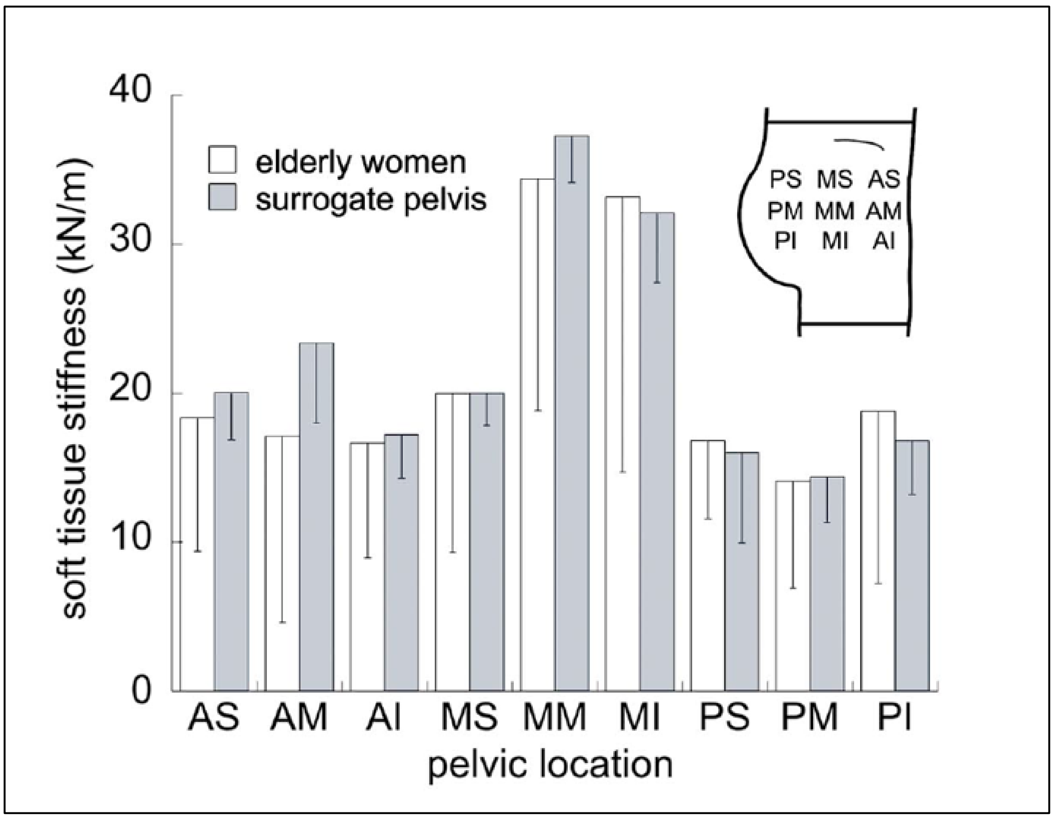

Figure 2. 12 Average soft tissue stiffness for elderly women (n=15) and the surrogate pelvis. Error bars show one SD. [13]

AS = anterior superior, AM = anterior middle, AI = anterior inferior, MS = middle superior, MM = center, MI = middle inferior, PS = posterior superior, PM = posterior middle, PI = posterior inferior

iii) Trochanteric Soft Tissue Thickness

Trochanteric soft tissue thickness is a measure of the lean and fat tissue that extends laterally from the greater trochanter [36]. The thickness of the human tissue above the greater trochanter varies according to individuals. Some studies report a thickness of 20 mm was defined for the layer above the most prominent part of the greater trochanter, corresponding to a typical thickness of soft tissue found for female hip-fracture patients [37]. Others report a thickness of the soft tissues overlying the greater trochanter was 2.4 cm [13]. A recent case-control study in postmenopausal women demonstrated that lower trochanteric soft tissue thickness was associated with greater risk of hip fracture and that lower trochanteric soft tissue thickness increased the estimated force applied to the proximal femur in a sideways fall and consequently increased the factor-of-risk as well [22].

iv) Hip Surface Geometry

To determine the surface geometry of the hip, a study where participants consisted of 15 Canadian women with a mean age of 77.5 years, mean body mass of 61.2 kg, mean height of 1.61 m, and mean body mass index of 23.6 kg/m^2. It reported 3D coordinates describing the average surface geometry of the hip, buttock and anterior thigh region of elderly women. The average width of the pelvis (left to right greater trochanter) was 362 mm and the average depth of the pelvis (sacrum to abdomen) was 266 mm [13]. In other studies, the hip surface geometry is derived from the circumference of outer thigh about the hip of a study into the anthropometric of elderly women [3, 38].

2.7 Hip Protector

The purpose of wearing a hip protector to prevent fractures seems reasonable in reducing the force exerted on the hip in a fall. In brief, a hip protector is a device that attenuates impact force that is placed over the greater trochanter to prevent hip fractures [39]. There are two types of hip protectors namely hard shell and soft shell hip protectors (Figure 2.13). The general mechanism of hip protectors are energy absorbing, energy shunting or both energy absorbing and energy shunting. Energy absorbing hip protectors are designed to attenuate impact forces by means of a soft shock absorbing material. On the other hand, energy-shunting devices distribute impact loads away from the greater trochanter to the surrounding soft tissues. There is a conflict in experimental findings whether soft or hard hip protector is better at attenuating the impact force as shown in Table 2.3 and Figure 2.14 [40]. Some devices combine both energy shunting and energy absorption into one product [7, 9, 41].

An example of a hard shell type is a hip protector made from polyurethane resin typically providing a preventive effect by energy shunting and a soft-shell type was made from polystyrene elastomer primarily providing a protective effect by energy absorption [27, 40]. There are also hip protectors fabricated with an outer rigid shell of fiberglass reinforced polymer composite and an inner layer of energy absorbing material [8]. Specifically, 'soft shell' protectors consisted primarily of foam and fabric, while 'hard shell' protectors contained a relatively stiff material that bridged over the greater trochanter [42].

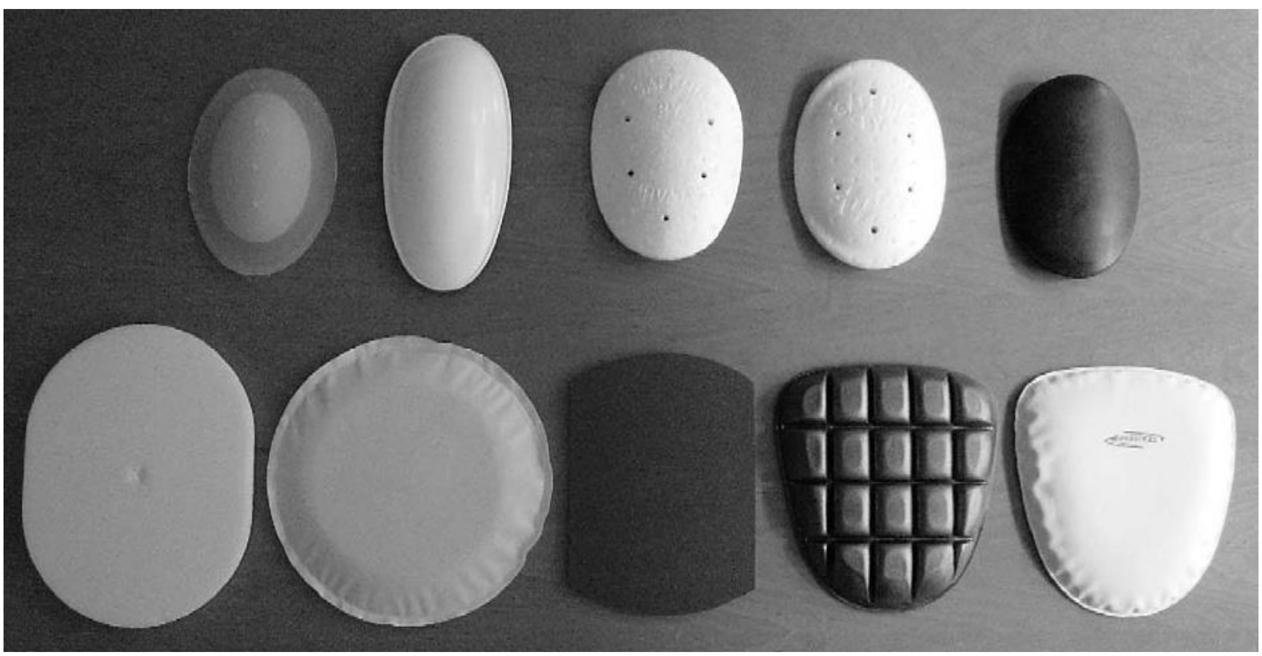

Figure 2. 13: Hip protectors. Top row hard hip protectors (from left to right): Hornsby healthy hip; KPH2, Safehip (old); Safehip (new); Impactwear Hip Protective garments. Bottom row soft hip protector (from left to right): Gerihip; HipSaver; Lyds Hip Protector; Safety Pants (FI); Safety Pants (NL) [27]

Table 2. 3: Force attenuation capacity of hip protectors [27]

Hip protector	1-inch soft tissue test	1/2-inch soft tissue test
	Mean ± SD in Newton	Mean ± SD in Newton
Calibration hit	7806 ± 69	7806 ± 69
Soft tissue	3998 ± 135	6378 ± 141
Hard hip protectors		
Hornsby Healthy Hip	854 ± 50 (−79%)	862 ± 29 (−86%)
KPH2	804 ± 22 (−80%)	900 ± 50 (−86%)
Safehip		
Old model	1298 ± 81 (−68%)	2061 ± 156 (−68%)
New model	911 ± 33 (−77%)	1817 ± 71 (−72%)
Impactwear Hip Protective Garments	971 ± 57 (−76%)	2105 ± 405 (−67%)
Soft hip protectors		
Gerihip	1957 ± 133 (−51%)	4948 ± 235 (−22%)
HipSaver	1689 ± 44 (−58%)	3472 ± 624 (−46%)
Lyds Hip Protector	1984 ± 73 (−50%)	4423 ± 66 (−31%)
Safety Pants (Finland)	2330 ± 67 (−42%)	5186 ± 129 (−19%)
Safety Pants (The Netherlands)	1520 ± 128 (−62%)	3415 ± 201 (−46%)

The average peak force of six tests, the standard deviation and the percentage attenuation as compared with the soft tissue hit are presented.

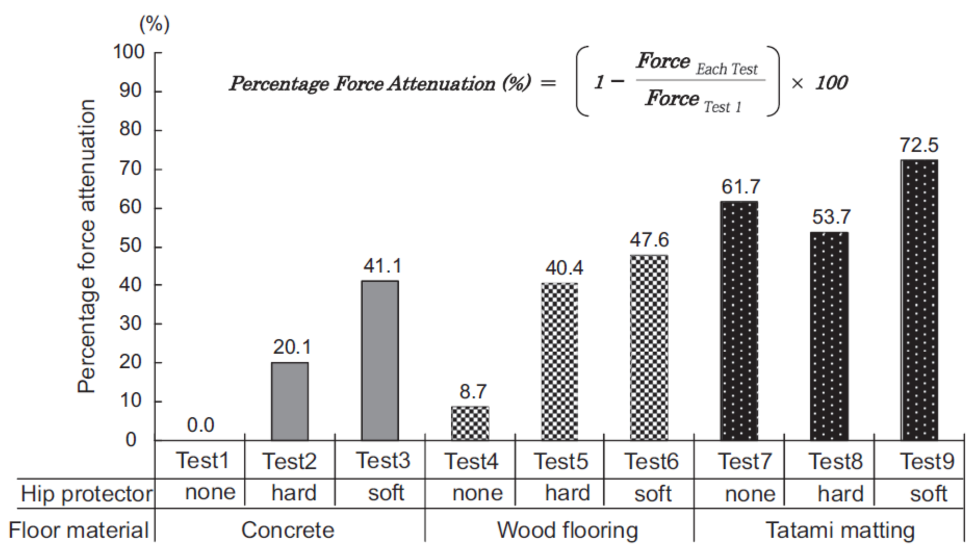

Fig. 6. Percentage force attenuations. Force attenuation percentages under each test condition were calculated relative to test 1 = 0.

Figure 2. 14: Percentage force attenuations. Force attenuation percentages under each test condition were calculated relative to 1 = 0 [40]

Despite numerous studies being conducted, the clinical effectiveness of hip protectors are still under debate [10, 11] and tends to depend on the severity of the fall and the compliance of the user [12, 13]. Use compliance can be seen as the willingness of a person to wear the hip protector [43]. It was approximated that an average of 50% of users do not comply in wearing the hip protector [6]. This is mainly because of the discomfort or the dislike of their appearance by the person wearing it. Improvements must be made in mobility and usability to overcome this problem [15]. The ergonomic design process such also consider the fitting of the hip protector to the user need to be considered. The hip protectors designed for western people would not be suitable for Asians people as their frame is smaller than western people [14]. The improper fit of the hip protector can reduce the effectiveness to prevent a hip fracture [15].

CHAPTER THREE
RESEARCH METHODOLOGY

3.1 Overview

This chapter highlights the methods used to conduct this research. It includes the simplified modelling of a femur bone, selection of material to represent the human soft tissue, surface geometry of the hip, fabrication of the soft and hard hip protector and the experimental setup of the drop impact tester. Each subsection explains in detail the methods used and illustrates the process taken in order to complete the task.

3.2 Fabrication of Femur Bone

Most hip fractures are due to a direct impact on the greater trochanter of the femur bone. A simplified model of the femur, specifically the profile of the greater trochanter was traced from a Sawbones 3rd and 4th generation computer aided design (CAD) model and modelled in SolidWorks 2016 (Figure 3.1). Focus was placed on the greater trochanter because hip protectors are designed to reduce peak force at the proximal femur (and fracture risk) by either decreasing the stiffness of the contact site (greater trochanter) or by forming a bridge over the trochanter to shunt the energy of the fall to surrounding regions where it can be absorbed more safely [6].

Aluminium was used to represent the femur bone. Aluminium is suitable because of its light weight, force transmission and it can withstand the high impact force of a typical sideway fall. For this experiment we have identified that a force of 2.5 kN is the is the threshold force for a hip fracture to occur [3, 6]. The aluminium block was milled to the proper dimension (195 × 40 × 50) mm and a 3-axis computer numerical control (CNC) machine (Robodrill α-T21iFLb) was used to machine the profile of the greater trochanter according to the design in the SolidWorks 2016 file (Figure 3.2). Two aluminium models of the profile of the greater trochanter was fabricated (Figure 3.3) in case of unwanted dents or defects in the aluminium profile of the greater trochanter if the impact force is too high.