Click or tap here to enter text.

MECHANICAL EVALUATION OF FEMUR GEOMETRY AND HIP PROTECTORS USING PENDULUM-BASED SYSTEM

By

AHMAD AMIR SHAFIQ BIN AHMAD IQBAL

(Matric No: 128923)

Supervisor

Dr. Mohamad Ikhwan Zaini Ridzwan

May 2019

This dissertation is submitted to

Universiti Sains Malaysia

As partial fulfilment of requirement to graduate with honours degree in

BACHELOR OF ENGINEERING (MECHANICAL ENGINEERING)

School of Mechanical Engineering

Engineering Campus

Universiti Sains Malaysia

DECLARATION FORM

This work has not previously been accepted in substance for any degree and is not being concurrently submitted in a candidate for any degree.

Signed ……………………………………… (AHMAD AMIR SHAFIQ BIN AHMAD IQBAL)

Date ………………………………………….

STATEMENT 1

This thesis is the result of my own investigation, except where otherwise stated. Other sources are acknowledged by giving explicit references. Bibliography/references are appended.

Signed ……………………………………… (AHMAD AMIR SHAFIQ BIN AHMAD IQBAL)

Date ………………………………………...

STATEMENT 2

I hereby give consent for my thesis, if accepted, to be for photocopying and for interlibrary loan, and for the title and summary to be made available outside organizations.

Signed ……………………………………… (AHMAD AMIR SHAFIQ BIN AHMAD IQBAL)

Date ………………………………………….

ACKNOWLEDGEMENT

First and foremost, I would like to express my utmost gratitude to my supervisor, Dr. Mohamad Ikhwan Zaini Ridzwan, lecturer of School of Mechanical Engineering, Universiti Sains Malaysia for guiding me from the beginning of this final year project. His willingness to spend time to advice on my progress has resulted in the completion of this project.

Next, I would like to thanks to the research officer, Encik Wan Mohd Amri Wan Mamat Ali for allowing me to use the load cell and all the experiment setup in the vibration lab. I would have not been able to complete my project without his assist while conducting the experiments.

I also extend my thanks to assistant engineers, Encik Faris Mohamed Zailan, Encik Abdul Halim Che Aat, Encik Baharom Awang. and Encik Norijaz Abd Aziz for their willingness to share their ideas that allowed me to improve the quality of my work.

I would like to thank Mr. Yahaya Suleiman Abimbola, for his deep knowledge and insight about this project that allowed me to understand my project better.

I am extremely grateful to my family for their love, prayers, caring and sacrifices for educating and preparing me for my future. With their support behind me, I was motivated to finish this project.

Finally, I would like to thank everyone who has been indirectly involved while completing my project. It has indeed been a very satisfying experience completing this project.

TABLE OF CONTENTS

LIST OF TABLES

Page

LIST OF FIGURES

LIST OF ABBREVIATIONS

CAD	Computer aided Design
3D	Three Dimension
Ac	Actual geometry of femur
CNC	Computer numerical control
DC	Direct current
Hp	Hard hip protector
IGES	Initial Graphics Exchange Specification
LSR	Load Strength Ratio
SFU	Simon Fraser University
Sm	Simplified geometry of femur
Sp	Soft hip protector
w/o p	Without protector
F_p	Peak force
ΔT	Total time impact
t_p	Peak force time impact

PENILAIAN MEKANIKAL TERHADAP GEOMETRI FEMUR DAN PERLINDUNGAN PINGGUL MENGGUNAKAN SISTEM BERASASKAN BANDUL

ABSTRAK

Keretakan pinggul akibat jatuh dari arah sisi adalah masalah kesihatan utama seluruh dunia yang menyebabkan kelumpuhan dan peningkatan risiko kematian pramasa terutamanya kepada golongan tua. Dalam kajian ini, pensimulasi impak pinggul berasaskan bandul telah dibangunkan untuk meniru situasi jatuh dari arah sisi. Untuk mensimulasikan impak lateral secara langsung, profil luaran trokanter besar telah digunakan dalam kajian sebelum ini untuk mewakili geometri femoral sebenar. Kajian ini direka untuk menyelidik kesan berpotensi terhadap ciri geometri femoral yang dikaitkan dengan puncak daya impak. Hasilnya menunjukkan bahawa puncak daya impak hanya berbeza 6% antara geometri sebenar dan dipermudahkan selepas penormalan berat bagi kedua-dua femur. Geometri femur sebenar mempunyai masa daya ke puncak yang lebih panjang iaitu 9.1% dan jumlah masa impak sehingga 12.12% berbanding dengan geometri femur yang dipermudahkan. Oleh itu, kajian ini menunjukkan bahawa geometri femur yang dipermudahkan boleh digunakan untuk mewakili geometri femur sebenar disebabkan oleh perbezaan daya impak yang kecil. Kajian ini juga menyelidik perbandingan biomekanik terhadap pelindung pinggul lembut (HipSaver) dan keras. Pelindung pinggul keras bercetak 3D direka untuk melengkapi geometri permukaan pinggul secara khusus. Keputusan menunjukkan bahawa pelindung pinggul lembut dapat meningkatkan masa ke puncak dari 21.72% hingga 26.67% dan jumlah masa impak dari 23.26% hingga 29.53% lebih baik daripada pelindung pinggul keras pada ketinggian impak tertinggi. Dengan ambang

keretakan pinggul sebanyak 5.2kN, pelindung pinggul lembut dapat mengurangkan 10% hingga 12% puncak daya impak berbanding pelindung pinggul keras. Dari segi keselamatan, hasil menunjukkan bahawa HipSaver lebih baik untuk mencegah keretakan pinggul daripada pelindung pinggul dicetak 3-D.

MECHANICAL EVALUATION OF FEMUR GEOMETRY AND HIP PROTECTORS USING PENDULUM-BASED SYSTEM

ABSTRACT

Hip fracture due to a sideways fall is a major health problem around the world which causes paralysis and an increased risk of premature death, especially to older people. In this study, a pendulum-based hip impact simulator was developed to mimic a sideways fall. To simulate a direct lateral impact, the external profile of the greater trochanter was used in the previous studies to represent the actual femoral geometry. This study was designed to investigate the potential effects of femoral geometric feature associated with the peak impact force. The result showed that the peak impact force varied only 6% between actual and simplified geometries after normalizing the weight for both femurs. The actual femur geometry has longer time rise to peak force up to 9.1% and total impact time up to 12.12% compared to the impact on simplified femur geometry. Therefore, the study suggested that the simplified femur geometry could be used to represent the actual femur geometry due to the small difference in impact force. The study also investigated the biomechanical comparison of soft (HipSaver) and hard hip protectors. The 3D-printed hard hip protector was designed to specifically complement the hip surface geometry. The results showed that the soft hip protector could increase the time to peak force from 21.72% to 26.67% and total impact time from 23.26% to 29.53% better than the hard hip protector at the highest impact height. With the hip fracture threshold of 5.2kN, the soft hip protector could reduce 10% to 12% peak impact force than the hard-hip protector. In term of safety, the results suggested that the HipSaver could be better to prevent hip fractures than the 3-D printed hip protector.

CHAPTER 1

INTRODUCTION

1.1 Background

Hip fracture is one of the health issues that give serious threat to the public, especially the elders and can cause paralysis or worst, death [1]. Hip fracture is the bone fracture that occurs at the proximal femur, at the outer area where the femoral head meets the acetabulum within the pelvis. There are three major types of hip fracture based on the anatomical site: femoral neck, intertrochanteric and subtrochanteric fracture [2]. The fracture mostly occurs as the result of a fall and impact on the greater trochanter of the femur. It can also be depicted that the probability of fall, the strength of the femur on the impacting side and the load applied to be the causes of the fracture [1].

In 1990, there are About 1.3 million hip fractures occurred worldwide; by 2025, this number is estimated to rise to 2.6 million and to 4.5 million by 2050, assume that there is no age-specific increase. Estimations that include an age-specific increase give estimated values of between 7.3 and 21.3 million by 2050[1]. In Malaysia, the incidence of hip fracture among of individual above 50 years of age was 90 over 10000 population [2]. Approximately 20% of older adults hospitalized for a hip fracture die within a year and about 50% will suffer a major decline in independence [3].

Hip fracture possibly occurs in many factors such as age, sex, and body mass index. Individual older than 70 years old has higher possibility to hip fracture compared to younger people due to bone mineral density (BMD) is already generally below the fracture threshold and the rate of bone loss has slowed[4]. Most of the elderly fallers who fractured a hip has a slow protective response of the body to break the fall with an outstretching the arms [5]. The threshold of the force that will cause a femur to fracture is 3.5kN [6].

The effective prevention of hip fractures can be achieved by the reduction of the number and harshness during falls. Studies were suggested that by using a protecting device such as hip protector can reduce the severity of the falls [7]. The hip protector can reduce the force applied to the proximal femur during the fall-related impact have the potential to reduce the hip fracture. There are two types of hip protectors, namely hard shell hip protector and soft shell hip protector [4]. Hip protectors with energy-absorbing or energy-shunting properties have been designed for active prevention of hip fracture [3]. However, they are lacking of market regulation, conflict in clinical value due to lack of agreement on techniques for measuring [1, 5, 8, 9] and optimizing the biomechanical performance of hip protectors [7, 10-12].

The purpose of this project is to develop the testing system that can accurately simulate the sideways fall. The experiment setup should be able to represent the conditions of a sideways fall on the greater trochanter which results in a hip fracture. The testing system can measure the force applied to the soft tissue that covers the hip region and the force impact to the femoral neck during a simulated sideways fall. The basic idea of the design for the testing system is based on the SFU hip impact simulator, the improved test system of Robinovitch et al [3].

1.2 Problem Statement

From the previous experimental studies, such as fall experiment by volunteers might be challenging, especially to older volunteers. By introducing this testing system, the impact can be simulated without the use of a volunteer. It's difficult to predict the impact force applied on greater trochanter during sideways fall. The force applies may differ depending on how the person fell. Some factors such as impact surface, the natural damping effect of body, fall direction and the person's weight will give a different impact. This project proposed to accurately simulate the impact of sideways fall.

Besides, there is various type of femur bone design such as the simplified geometry of femur bone and actual geometry from the previous experiment for hip impact test. The different type of femur bone geometry may vary the impact force and force applied to the femur bone. These problems will be validated thru experimental testing using different geometry of femur bone.

The efficacy of hip protectors still on the debate although many studies had been conducted to prove the effectiveness of hip protectors. This project will be validated the effectiveness of the market hip protectors and designed hip protectors thru experimental testing.

1.3 Research Objectives

The objective of this research is:

1. To analyses the difference impact force of a sideways fall between simplified and actual geometry of femur bone.

2. To evaluate the impact force with soft hip protector and hard hip protector.

1.4 Scope of Work

This project involves design, fabrication, experimentation, and analysis of the result. First, the hip impact simulator testing system had been designed based on the previous testing system. The system is to measure the impact force applied to the hip during a sideways fall from standing height. The fabrication of this testing system started with the material selection of each part of the simulator. The crucial part for this simulator is the leaf spring, to have similar effective stiffness of the pelvis during the impact on the hip. Due to the difficulty in finding material for the leaf spring, these part in exclude for this experiment.

Besides, the surrogate pelvis model such as the proximal femur and soft tissue is another part that is important for this research. The surrogate pelvis must have exact properties as actual human hip to perform the biomechanical testing of the hip protector. The material selected for the femur bone is aluminum and the geometry are according to the actual human femur bone. For the soft tissue, a different type of densities of polyethylene foam is used which has similar properties to human tissue. The surrogate pelvis was placed on the impact pendulum to be tested with and without the hip protector.

1.5 Thesis Outline

This thesis consists of five chapters. An overview of the factor occurrences of hip fracture among the elderly and the way to prevent it to happen which is by using a hip protector. This chapter briefly explained the objective of this project and the scope of work which this project will cover. Chapter two determines the method to run the experiment and the raw material that is suitable to use in this project. Chapter three explained the important stages of completing this project such as fabrication process and the experimental setup. The analysis of the impact testing result is discussed briefly in chapter four. Lastly, chapter five discussed the overall finding, results and the future recommendations for this project.

CHAPTER 2

LITERATURE REVIEW

2.1 Overview

In this section, the anatomy of the hip described clearly. The different type of experiment on the biomechanical test of the hip protectors was reviewed to evaluate the suitability of experiment rig used to test the hip protector. The impact force and femoral strength to cause the hip to fracture were also identified. Lastly, the effectiveness of the pelvic stiffness was identified to simulate the stiffness of actual human pelvis.

2.2 Anatomy of hip

The hip is ball-and-socket joint bounded by strong and well-proportioned muscles[13]. The hip joint provides stability, allowing a wide range of movement in some physical planes and involved in transferring the body weight [14]. The hip consists of four characteristics of a joint cavity (synovial or diarthrodial joint), joint surfaces are covered with articular cartilage, it has a synovial membrane producing synovial fluid, and surrounded by a ligamentous capsule[15]. The structures of the femur bone shown in Figure 2.1. Figure 2.2 shows the anatomy of the hip joint.

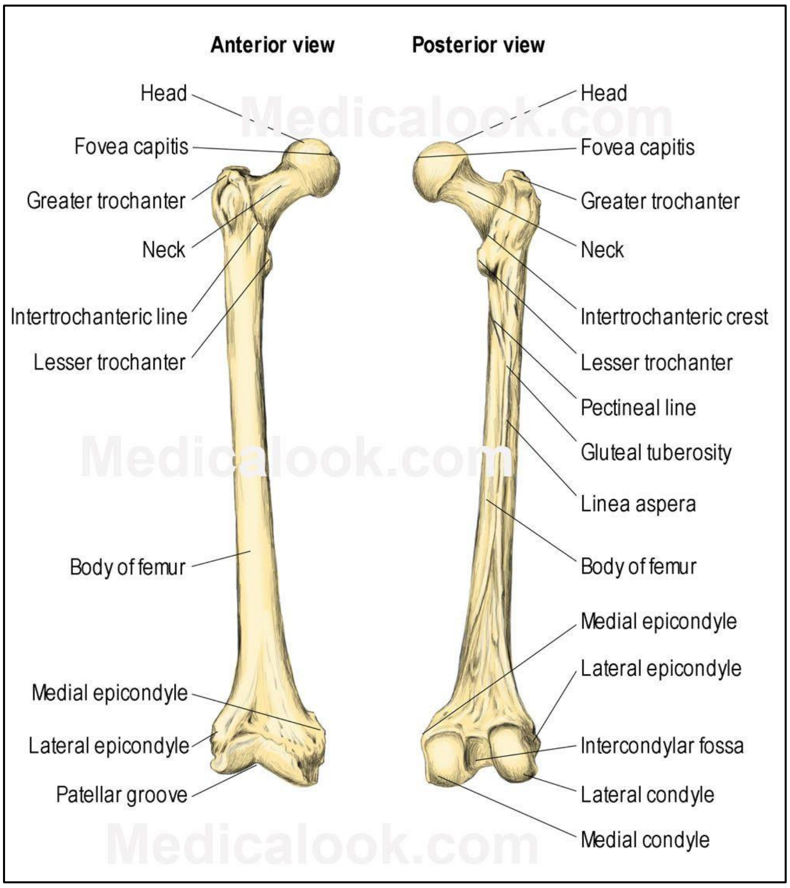


Figure 2.1: The anatomy of the femur bone in anterior and posterior view (www.pinterest.com)

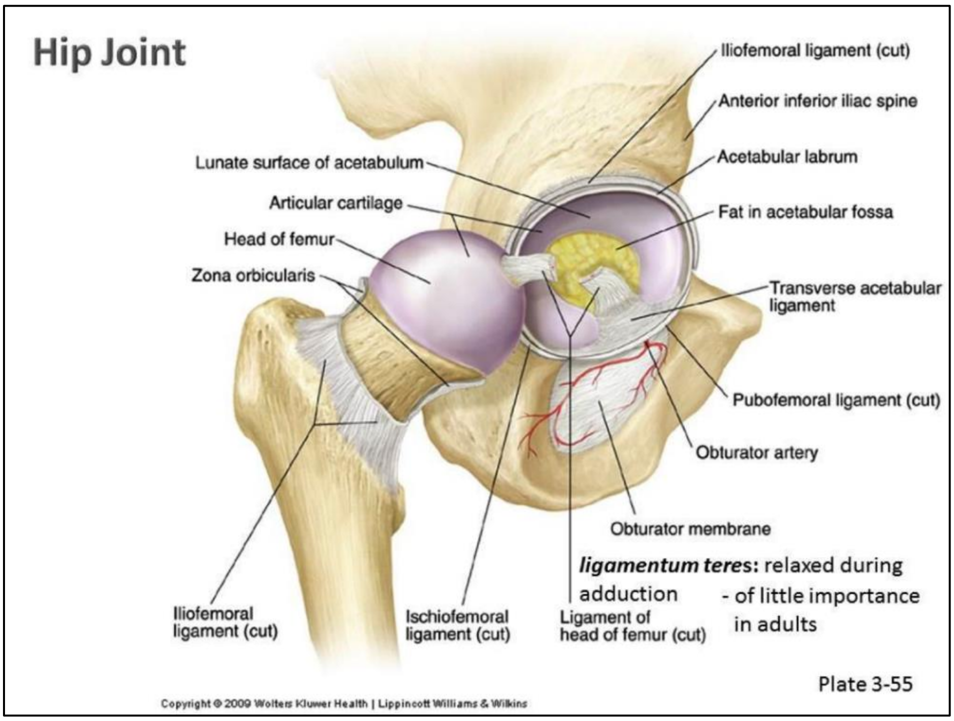


Figure 2.2: The anatomy of hip joint (www.surreyosteopathiccare.com)

2.3 Biomechanics Testing

The fall simulator consists of two stages of the experiment. The first stage is the fall phase and the impact is by a gravity-driven inverted pendulum-style fall with only one rotational degree of freedom about an axis through the foot point to simulate the protective fall to the side [16, 17]. The second stage is the impact phase where the initial conditions such as velocity, alignment, and an unconstrained impact are controlled [16]. Pelvic cadaveric specimens are embedded in surrogate soft tissue and attached to the lower limb construction that designed to mimic the actual thighs and calves [17]. This experimental setup is shown in Figure 2.3.

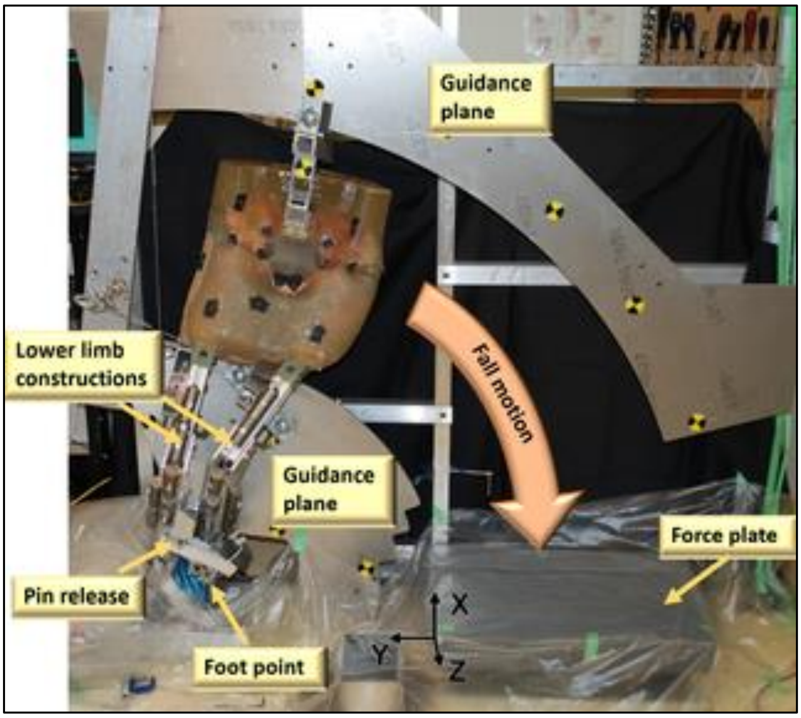


Figure 2.3: Fall simulator [16]

Laing, et. al.,2009, SFU hip impact simulator (Figure 2.4) is the test system that had been improved from the test system of Robinovitch et al as shown in figure 5 [3, 12]. SFU hip impact simulator consists of surrogate pelvis connected to the impact pendulum via leaf spring that simulates the total effective stiffness of the pelvic. The surrogate pelvis that contain the combination of simulated soft tissue and proximal femur, was designed to simulate the actual surface geometry and local variation of the elderly patient soft tissue. The test system is released by electromagnet from a certain angle in the incline position and then hit the ground in a horizontal position. The applied force on the femoral neck is measure using load cell placed on it while the impact force on the skin surface is measured with a floor-mounted force plate [3, 4].

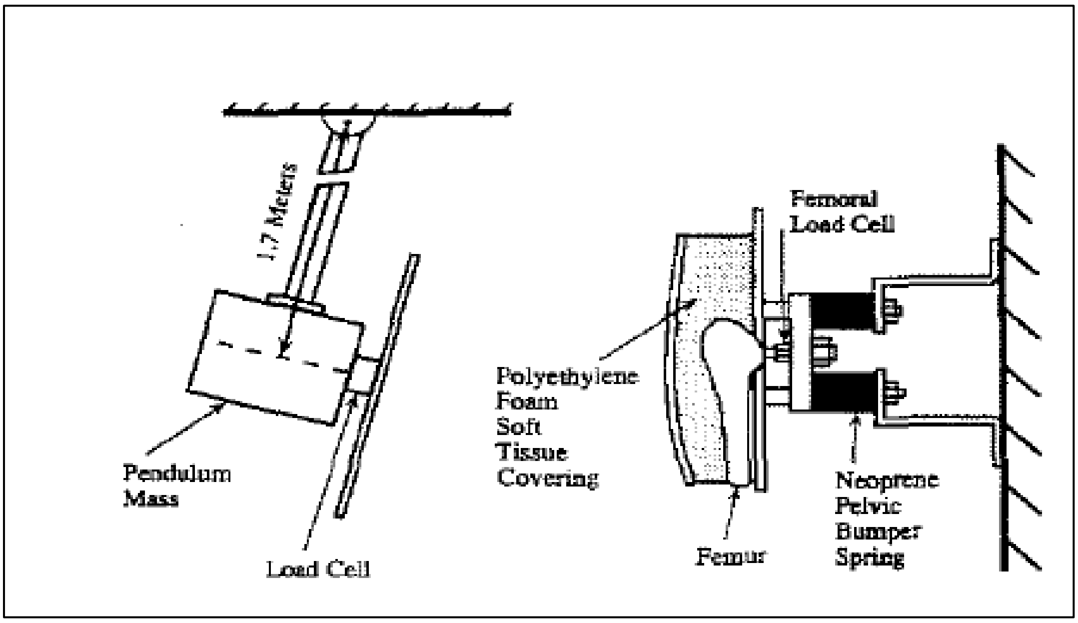


Figure 2.4: Surrogate pelvis and impact pendulum [12]

Other experimental studies such as voluntary natural fall from standing height by young adults, pelvis-release experiment to measure the damping properties of hip soft tissues and predict the impact force (Figure 2.5) [18], and fall from a kneeling position to measure the impact force in a low-severity fall have been conducted [19]. Dynamic models have been developed to determine the fall-induced impact force [6].

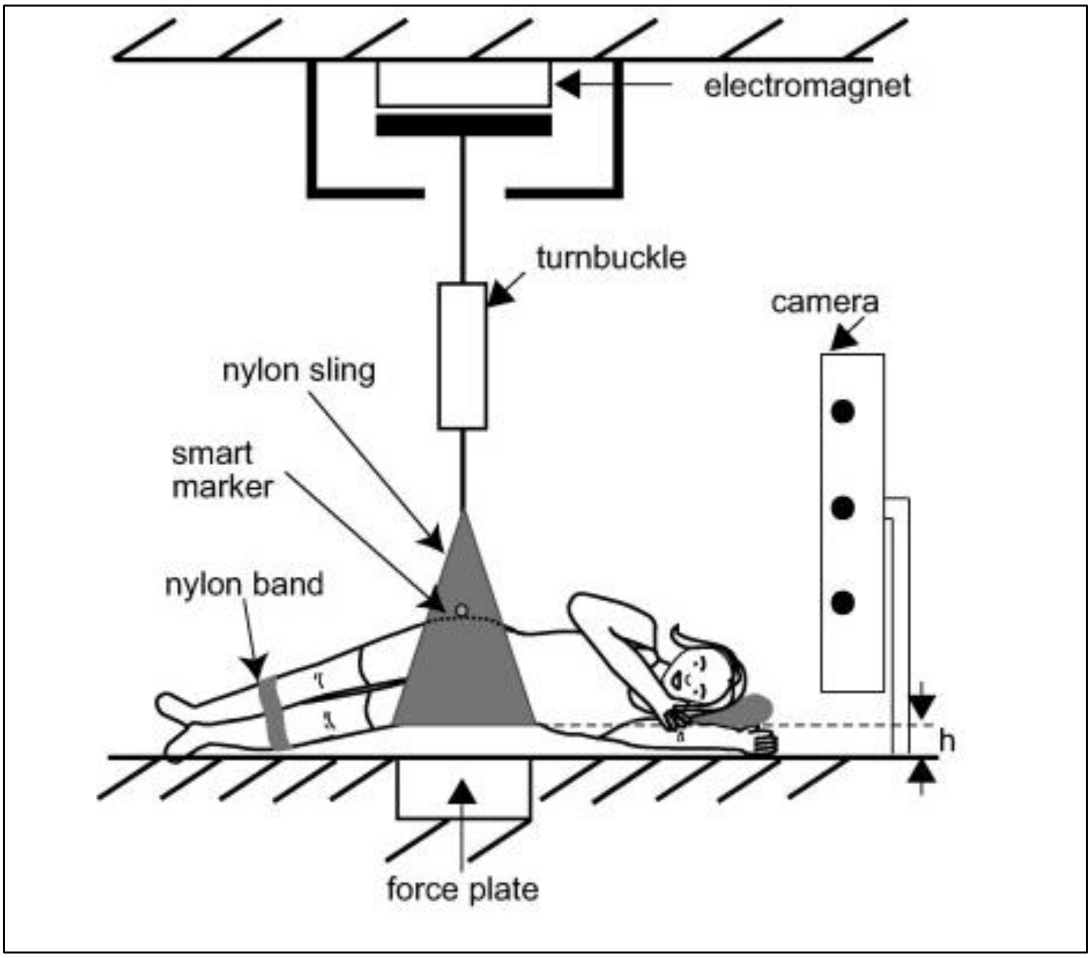


Figure 2.5: Schematic of equipment and participant positioning during the lateral pelvis release experiment [18]

2.4 Determination of Impact Force

All test systems use a falling mass to generate the impact energy of a sideways fall on the hip. There are three common experimental studies that methods in which 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 and fall from a kneeling position to measure the impact force in a low-severity fall.

Sarvi et al. 2017, state that the impact velocities of a person can be up to approximately 5m/s in an unexpected sideways fall from standing height[20]. However, the average impact velocity to the hip is 3.0m/s with a standard deviation of 1.0 m/s [21]. The range of peak impact force in a fall from standing height is ranged from 4050 to 6420N [12]. The peak impact force can be increased up to 8600N for an average individual in a fall from a pelvis height of 0.7 m [22]. For a low-severity fall, the impact velocity is approximately 1.0m/s and the impact force can vary from 1 to 2.5kN. The mean values for an unexpected fall from sideways for an average individual impact velocity of 3.0 m/s (SD=1) and femoral force of 5200 N. Hence, Table 2.1 show the listed results of the range of the impact velocities and forces that impact on the hip in a fall from standing height.

Table 2.1: The result from studies reporting the range of impact velocity and force on the hip

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 Determination of Femoral strength

The effectiveness of a hip protector can be determined with the reduction of peak force at the proximal femur during the simulated fall impact below the value that will cause the fracture to the femur. Different value of force required to fracture the femur depending on the direction of the force acting on the femur. However, only the force acting on the greater trochanter is considered in this project.

A study from Courtney et al. 1995, found that the mean fracture force for the older was 3440 N, 7200 N for young adult. The mechanical testing is on proximal femur of 8 older individuals with mean age of 74 years and 9 younger individuals with a mean age of 33 years using fall loading configuration (Figure 2.6).

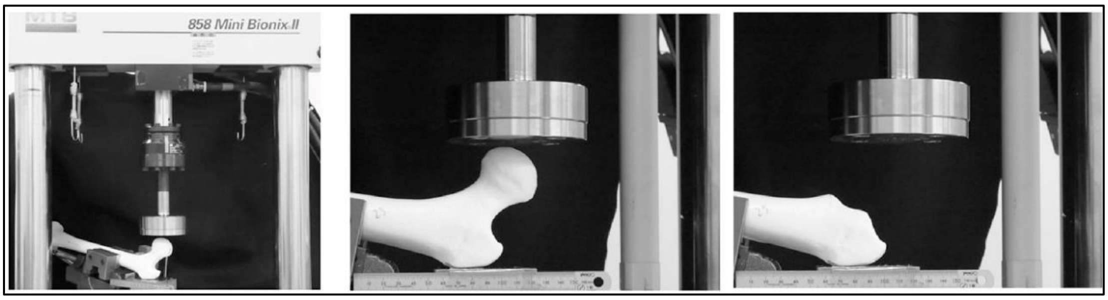


Figure 2.6: Biomechanical test on the proximal femur using simulated fall configuration[23]

Based on the study conducted by Robinovitch et al. 2009, analyzed from the result of 16 (Table 2.2) studies that have reported the strength of elderly proximal femur tested in a fall loading configuration. Femoral strength is defined as the compressive force either measured at greater trochanter or femoral neck that caused the fracture[24-31]. The data specify that age and gender have a significant effect on femoral strength. The median femoral strength 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). Besides, studies that reported age-specific values [29], 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 for 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 femoral strength of the cadaveric proximal femur from older adults in a sideways fall loading configuration

		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[c]				2,400[a]			70(52-92)[b]; 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.

Sarvi et al. 2017, found that the hip fracture risk is depending on the bone strength and the applied force on the hip based on the biomechanical point of view. The load and the strength ratio (LSR) [6, 32, 33] is measured to determine the hip fracture as the following equation:

$$\text{LSR}=\frac{The\ applied\ load}{The\ bone\ strength}$$

Where, $The\ bone\ strength$ = Maximum force that the bone can withstand without fracture

$The\ applied\ load$ = Fall-induced impact force on the hip

The bone will fracture if the applied load is higher than the bone strength (LSR>1) [6]. For this standard, the applied impact force to the femur is one of the two main factors of fracture risk[34]. therefore, there are many studies have been conducted to determine the range of the sideways fall impact force that can cause a hip to fracture.

There are two parameters are investigated, the range of the fall-induced impact force to the hip and the range of the force that can cause a fracture in a femur[6]. Figure 2.7 shows the summary for the range of the hip impact velocities (Figure 2.1) and fall impact force tables and the femur strength and the range of the force that can cause the femur fracture table (Figure 2.2) result in the range of the impact force and the average of the femur strength. Based on Figure 2.7, the median value of the impact force of unexpected fall from sideways for an average individual is 5.2kN and the average strength of the femur or fracture femur force is 3.5kN [6].

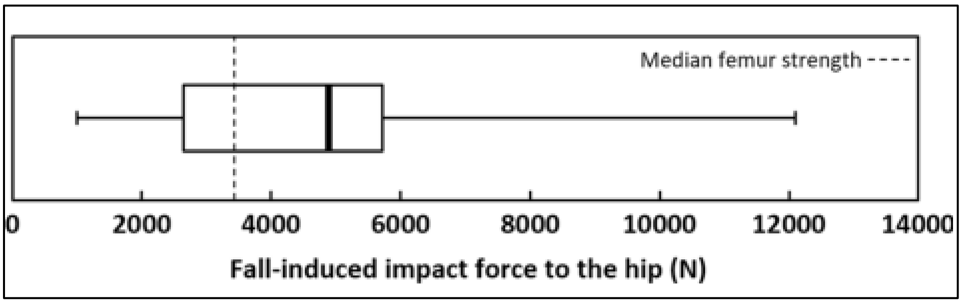


Figure 2.7: Box plot showing median, quartiles, and range of lateral fall impact force to the hip in comparison with the median femur strength[6]

2.6 Effective stiffness of pelvic

Robinovitch et. al., 2008, the stiffness of the pelvis is simulated using leaf spring that connected between the surrogate pelvic and impacts pendulum. The total effective stiffness produces by pelvic is 42.2kN/m. The total effective stiffness is the simulation of articulations between the pelvis, trunk, lower extremities, and the stiffness of the pelvis itself does not include with the stiffness of soft tissue [3]

Robinovitch et. al., 1997, the model of the Pelvis-release experiments is capable to simulate both the flexural and compressive deflections of the body during the experiment. The system consists of a single effective mass attached to three sets of spring-damper elements (Figure 2.8). The combined flexural stiffness, damping of the muscles and ligaments connections between the trunk, pelvis, and lower extremities representing by k_f and b_f.These elements constrain the hip and pelvis from lateral excursions from the midline of the body. Values for k_f , b_f and by for each subject are given in Table 2.2.

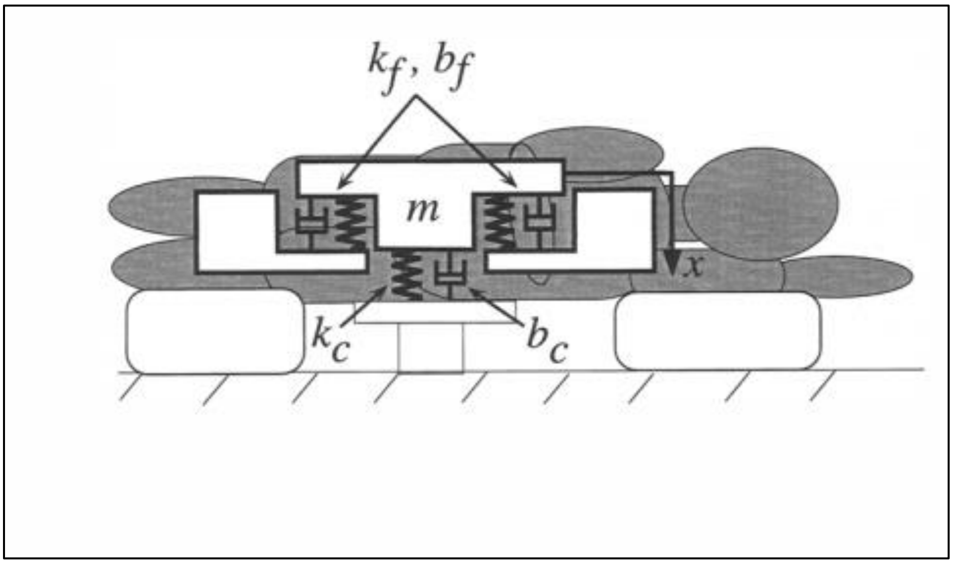


Figure 2.8: Mathematical model of the human body [35]

Table 2.3: Parameter values and predicted impact forces for each subject [35]

Subject	Body Mass (kg)	m (kg)[a]	k_c (kN/m)[a]	k_f (kN/m)[a]	b_c (N-s/m)[a]	b_f (N-s/m)[a]	F_c (N)[b]	F_f (N)[b]
		59.0	70.0	16.6	1380	219	4920	983
m1	82.8	29.2	46.8	8.3	899	89	2970	387
		35.7	49.7	6.4	292	318	2970	615
		26.2	23.7	6.3	187	146	1830	557
m2	61.2	15.9	13.9	4.7	215	130	1150	309
		32.3	62.2	5.0	312	225	3470	421
		41.4	45.8	9.8	1070	58	3560	440
m3	81.2	25.9	22.3	5.1	287	54	1850	409
		38.6	51.4	3.4	13	65	3670	272
		40.1	46.3	11.2	175	175	3180	752
m4	71.3	37.4	32.6	9.0	711	91	2680	573
		57.5	59.0	10.9	629	249	4360	932
		41.6	64.2	7.8	1059	168	1960	545
m5	83.9	45.8	50.9	8.8	703	247	3530	858
		46.9	66.0	3.7	364	158	5288	297
		26.2	29.0	5.5	439	31	2140	341
f1	63.0	32.7	36.0	7.3	609	38	2660	426
		42.0	65.4	8.3	320	256	4530	625
		27.7	27.3	4.3	522	91	2100	342
f2	61.3	51.5	48.9	8.7	1233	105	4100	509
		34.9	43.2	2.8	664	121	2990	298
		34.2	34.2	6.7	421	53	2660	490
f3	59.0	50.1	55.1	11.3	889	156	3950	776
		48.5	60.1	7.6	529	135	4250	597
		24.2	34.4	4.6	230	132	2210	375
f4	58.2	20.1	30.0	4.4	224	136	1820	357
		34.7	61.6	7.7	13	98	4210	535
		24.4	27.3	4.3	324	58	1992	321
f5	57.2	19.2	17.0	3.6	207	47	1452	271
		16.4	17.5	2.0	56	27	1503	181

[a]Cell entries correspond to values for configurations 1, 2, and 3, respectively.
[b]Impact forces are based on an impact velocity of 3 m/sec.

CHAPTER 3

METHODOLOGY

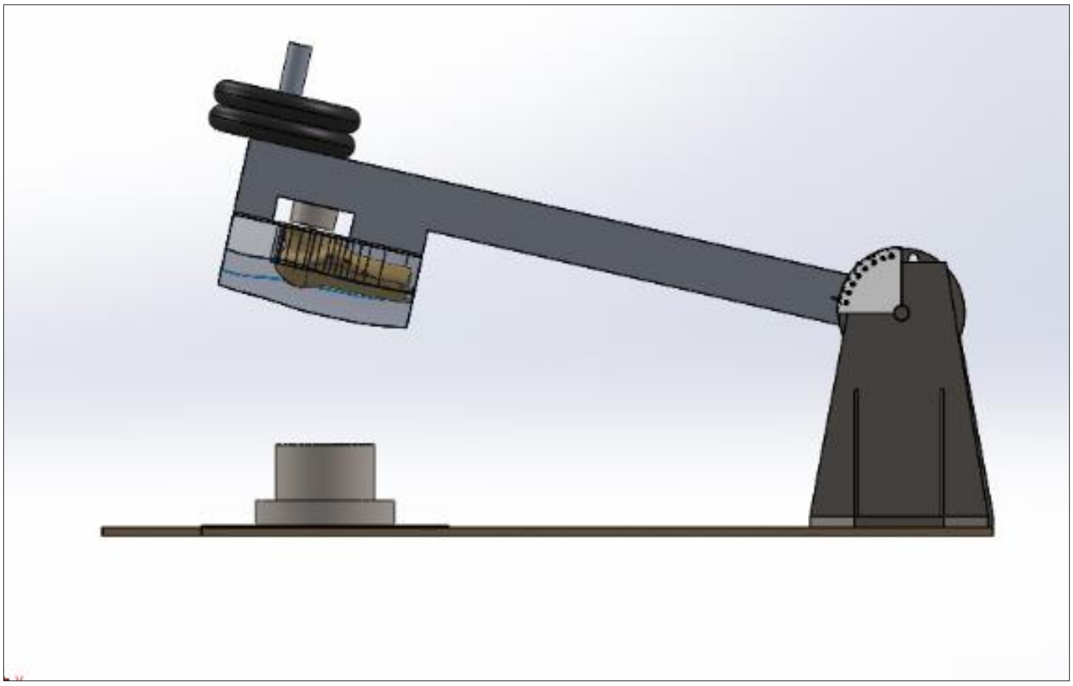
3.1 Overview

This chapter focus on the methods used to handle this project. It includes the fabrication of impact simulator, the actual geometry of a femur bone, simulated human tissue and the experimental setup of the hip impact simulator. The method used, and the process taken by this project is explained in each subsection.

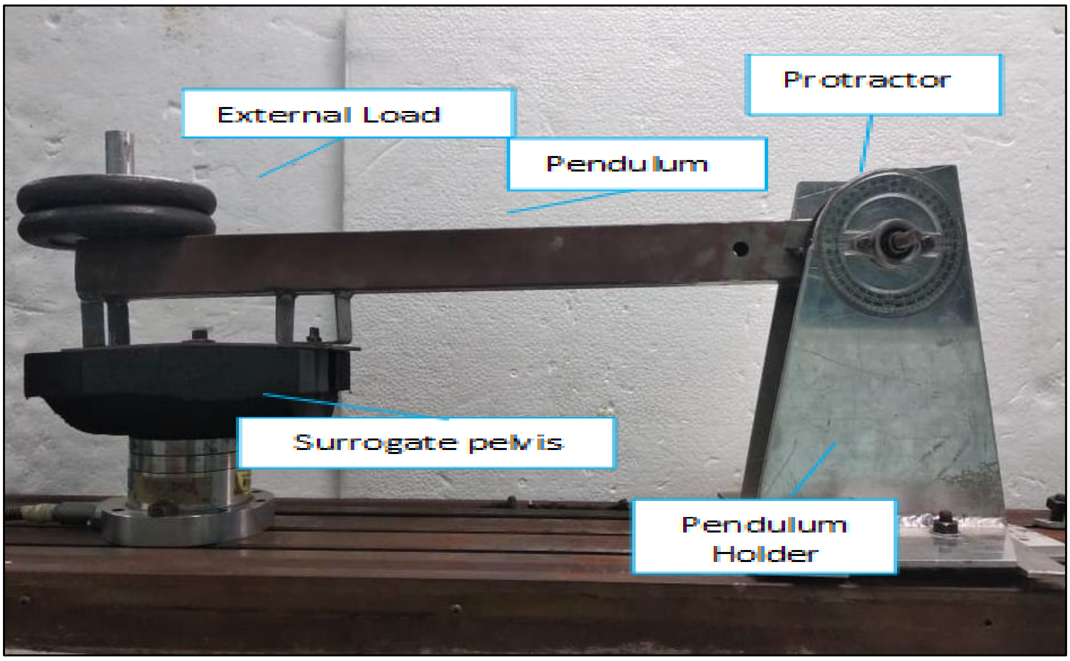
3.2 Fabrication of hip Impact Simulator

The general idea of hip impact simulator testing system was based on the previous study where the system is to measure the impact force applied to the hip during a sideways fall from standing height[4, 36]. This device comprises of the surrogate pelvis that's attached to the impact pendulum released from an inclined position then hit the ground in a horizontal position. The surrogate pelvis comprised of simulated human soft tissue and proximal femur[3]. Each part of the hip impact simulator is modelled in SolidWorks 2016 (Figure 3.1(a))

A pendulum impact with 70-cm length with adjustable slotted mass (2.5-5) kg with 2.5 kg increment was designed similar to the previous simulation device[36]. This simulator also completes with a protractor that attached to the pendulum holder to indicate the impact angle. A complete hip impact simulator (APPENDIX A1) testing system is shown in Figure 3.1(b).



(a)

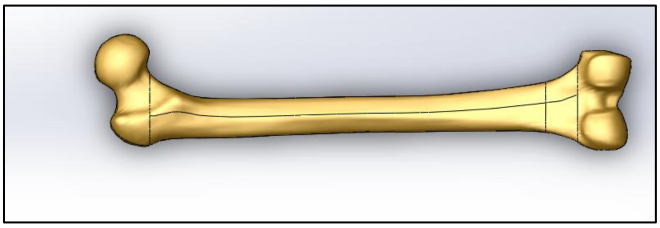


(b)

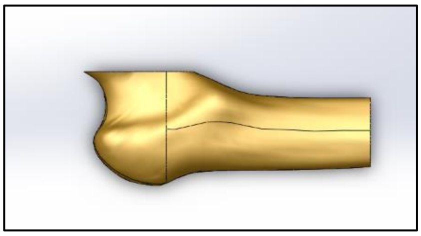
Figure 3.1: (a)The Hip impact simulator CAD Model (b)Completed Hip impact simulator model system

3.3 Fabrication of Actual Geometry of Femur Bone

The CAD model of the femur bone is based on the Sawbones shown in Figure 3.2(a). The proximal femur bone is used aluminum because the material is light, force transmission and it can resist the high impact force through the impact test. Due to the limitation of the Aluminum material, the CAD model of the femur bone was redesign as shown in Figure 3.2(b). The femur bone CAD design in SolidWorks 2016 file need to change into the Initial Graphics Exchange Specification (IGES) file because the 5-axis milling machine used Autodesk Fusion 360 software to operate the machine. The aluminum cylinder block was milled using a 5-axis computer numerical control (CNC) milling machine (DMU 40 Monoblock) to the desired actual geometry of femur bone according to the design in the SolidWorks 2016 file. The actual geometry of femur bone was fabricated shown in Figure 3.3. The comparison of actual and simplified femur geometry shows in Figure 3.3(b).

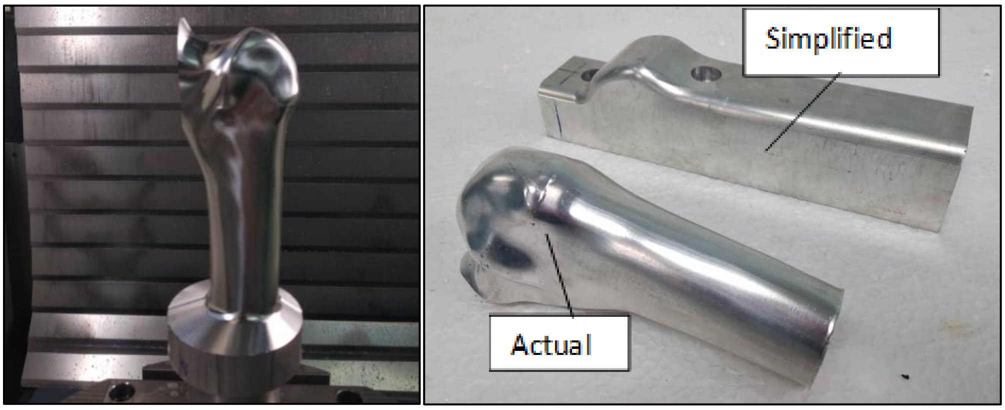


(a)



(b)

Figure 3.2: (a)Femur bone based on Sawbones CAD model (b)Actual Geometry of Femur Bone CAD model



(a) (b)

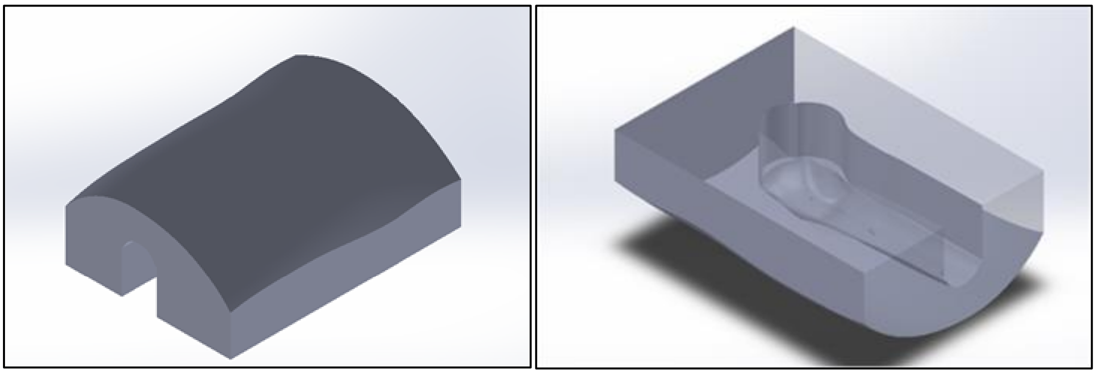
Figure 3.3: (a) Complete aluminum of actual geometry femur bone (b) Comparison of actual and simplified femur geometry

3.4 Fabrication of Simulated Human Tissue

The surface geometry of the hip was taken from 15 volunteers of Canadian women with a mean age of 77.5 years, mean body mass of 61.2kg, mean height of 161m and mean body mass index of 23.6 kg/m^2[3]. The SolidWorks 2016 CAD model of simulated human tissue was taken from the previous study where from 3 dimensional (3D) coordinates are plotted in Solidworks 2016 and the surface geometry of hip was formed[37]. The 3D coordinates relating the average of pelvic surface geometry of 15 women volunteer[3]. Modification of the cavity under the hip surface had been done to fit in the actual geometry of aluminum femur bone (Figure 3.4(a)).

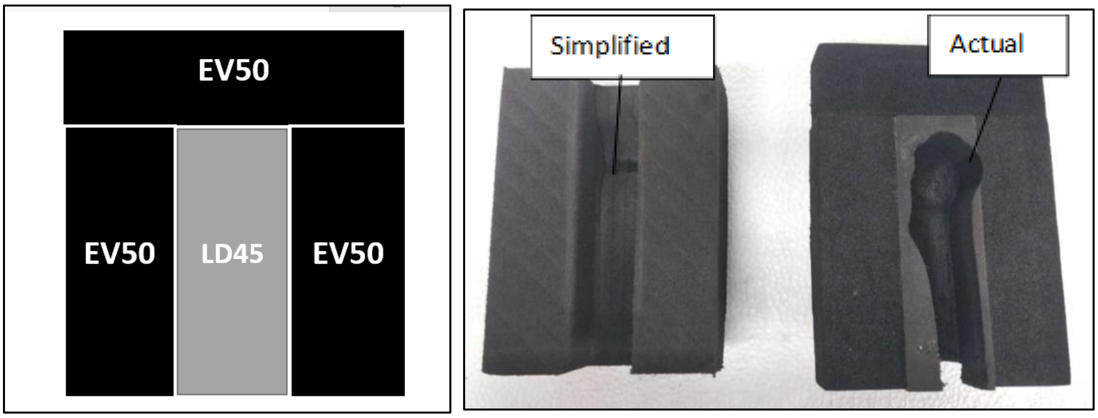
Closed-cell polyethylene foam is used to simulate the human tissue closed-cell. The closed-cell polyethylene foams Plastazote LD45 of density 45 kg/m^3 (APPENDIX A2) was used directly over the proximal femur and closed-cell copolymer foam Evazote EV50 of density 50 kg/m^3 (APPENDIX A3) was over the regions anterior, posterior, and superior to the femur[3]. This material is obtained from Wansern Technology Sdn. Bhd.

The polyethylene foam LD45 and EV50 are glued together using a spray adhesive (Spray idea 92) to form a single (220 x 150 x 90) mm³ as shown in Figure 3.5(a). The 3-axis CNC machine (Robodrill α-T2liFLb) was used to machine the block according to the SolidWorks 2016 file. The completed machined actual femur geometry soft tissue was compared to the simplified femur geometry soft tissue (Figure 3.5(b)).



(a) (b)

Figure 3.4: (a)SolidWorks 2016 CAD simulated hip model isometric view (b) Transparent isometric view of modification on the hip CAD model

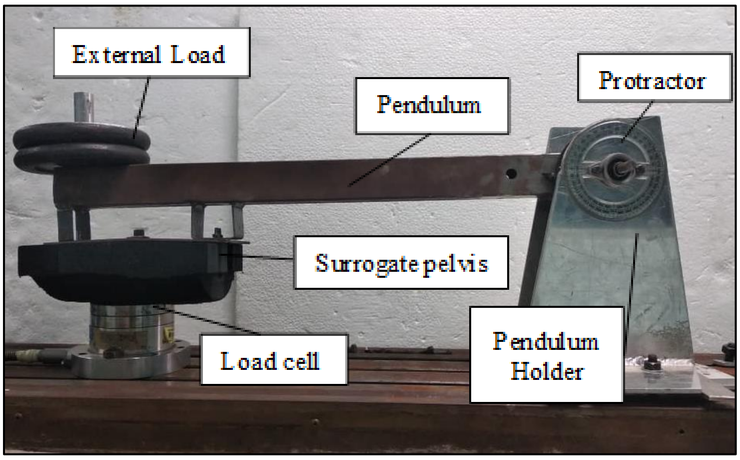


(a) (b)

Figure 3.5: (a)Schematic of polyethylene foam block (b)Comparison between complete machined actual femur geometry soft tissue and simplified femur geometry soft tissue

3.5 Experiment setup with the Hip Impact Simulator

The experiment was carried out in Vibration Lab at Mechanical Engineering School using hip impact simulator (Figure 3.6(a)). The simulator used to measure the impact force on the hip and measure the force applied to the femoral neck. The total impact force is measured with a load cell (Kistler Model CH-8408 Winterthur, Switzerland) placed at floor position. A load cell was located between the proximal femur and based plate and the force applied to the femoral neck is measured by a load cell (CAS Model MNC-200L. A 20kN load cell (Kistler) was connected to a Multichannel Charge Amplifier (Type 5070) and IMC device that connected to the desktop (Figure 3.6(b)). The impact force was received by the load cell and the data shown in the IMC software was recorded. For a 2kN load cell (CAS), it was connected to the 5-volt DC power supply and digital indicator (OMRON Model K3HB-V). Calibration had been done to this 2kN load cell with a 20kN load (Figure 3.7). The force applied to the femoral neck was recorded to the digital indicator in term of maximum force received.



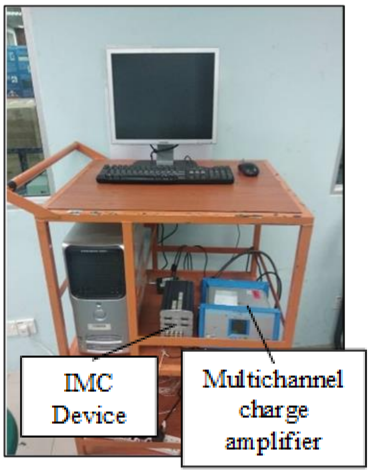


Figure 3.6: (a)The hip impact simulator testing system (b) The Multichannel charge amplifier, IMC device and desktop used to record the data received from load cell (Kistler)