

# Parametric analysis for the design of hip joint replacement simulators

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**Abstract:** The simulation of wear, between the components of artificial hip joint implants, is a complicated problem that does not have a robust analytical answer yet. Many studies have been conducted to predict the wear between the femur head and the acetabular cup, as the debris generated due to the wear might produce adverse effects after the surgery. Hip joint simulators provide a means to quantify the amount of wear in preclinical settings, as an in vitro method. However, this brings some other challenges in terms of bio-fidelity. The simulators use force and range of motion data as input and provide wear information as an output. For this reason, it is important to be able to simulate the realistic conditions, by the proper transmission of force and position controlling of the components. Many studies performed on wear simulators but none of them worked on the machine parameters such as power consumption and sensitivity to external inputs in detail. In this study, we perform a sensitivity analysis of the factors affecting the forces acting on the femur head. In silico simulations were performed by changing the values of acting force, friction coefficient, and radius of femur head to understand the effects of each parameter on the frictional moment of the joint. These analyses demonstrate the importance of using correct parameters while designing simulators, which accept flexible boundary conditions. The architecture of the hip simulator was also investigated for the first time. The results are expected to pave the way for improving the bio-fidelity of the simulators in the field of biomechanics.

**Keywords:** Total hip replacement, In vitro, Parametric design, Simulator

## INTRODUCTION

Total hip replacement (THR) is a widespread practice performed by surgeons, in which natural joints are replaced by artificial joints when joints lose their functionality. Artificial THR is an engineering solution to improve the quality of life of hip patients. The first hip arthroplasty was performed in 1962 by John Charnley a British orthopedic surgeon [1]. The design and improvement of hip joint replacement go back to the studies by Galilei [2] to identify the load-bearing bones. Later, Carlet and Marey determined the forces between the foot and the ground in a full “gait cycle” [3], [4]. In the studies of Quénu and Demeny, a measuring device called dynamograph was designed. The measurement is performed by placing an airbag beneath the heel and forefoot. By measuring the pressure difference between the two parts, a linear approximation of the load distribution is calculated. The measurement during walking activity produced the gait load cycle [5]. Although the shape of the load vs the time, a measurement performed by them is as in the current electronic devices, the load magnitude even for harsh cases such as running, does not exceed more than 20kg above the body

mass. In some cases, vertical loads show values less than body weight. The authors believed that there is a relation between this vertical load and the body’s center of gravity acceleration.

A device called basograph was invented [6], as in the design of Demeny and Quénu, and then further developments on the pneumograph were made to record such forces [7]. The pneumograph was a specific shoe measuring the pressure distribution on the foot palm. Further trials with Schwartz and his team were performed to introduce electrography for measuring gait cycle forces [8]. Such research using newly introduced methods such as force plate and motion cameras continued later [9], [10]. Apart from measuring the forces on the foot palm, many studies have been performed on measuring forces on the femur joint itself. The hip joint is reinforced by four ligaments and, most of the muscle forces [11], therefore, it is essential to have a deep knowledge about the effects of the reinforcement tissues to estimate correctly about the femur headloads. Culmann, under the influence of a presented paper by Meyer [12] estimated femoral loads to be about 30kg [13] which was not accepted by the majority since the effects of surrounding tissues were not accounted for. Later many other studies on muscles and tissue effects have been performed [14]–[16].

Using in vivo measurement is a method to understand the forces and reactions without losing data. Raydell designed a total hip replacement [17] implanted with strain gauges to measure the gait cycle dynamically due to the previous studies [18]. Such studies continued later using more advanced data acquisition devices [19], [20].

Simulators are planned as a good substitute for in vivo measurements. The use of simulators helps the scientists consider the effects of the forces and actions with more care. The first in-vitro hip joint replacement simulator [21] was a machine with three degrees of freedom performing measurements. The later design [22] has a rotation degree of freedom around the flexion and extension axis and could model vertical and sidewalk loads. The designers believed that due to their research, the vertical load has a prominent role in walking. The third design was performed by Walker and Gold [23]. This design aimed to calculate friction forces in the designed prostheses. All these designs tried to comply with the Raydell [17] in vivo force measurement, so they lost the synchronization between the force and the angle of the femur. Other designs considered synchronization in their flexion-extension motion simulator to measure the wear and friction [24]. The lubrication inside the joint and its effects on the femur head was also investigated [25]. Swanson et al. design their machine to test new materials for Hip and Knee joints [26]. Ungethüm et al. improved Dowson designs and

performed new curves [27]. By focusing more on wear, Beutler, Lehmann, and Stahli [28] designed a universal machine for different body joints. A new three-degree freedom machine was designed, and research was performed by Lewis Research Center, NASA to investigate the effect of femur head-on wear [29]. The machine could mimic all three rotational degrees of rotation in the hip joint. Introducing a new control method using a primary computer, Cappozzo and his colleagues [30] designed a control schematic for the hip simulators. The method was later improved by other researchers [31], [32]. Saikko designed a series of hip joint simulators for different measurements of wear and friction called HUT-1, HUT-2, HUT-3, and HUT-4 [33]–[36]. Many recent designs tried to perform the test on more than one femur head since tests on more than one station would provide means for statistical sampling and hence be more economic. The improvement of hip simulators in recent years focused more on the control side and introducing new phenomena to the machine design.

One of the major problems in hip joint implants is the friction between the artificial femur head and the acetabular cup. The wear and friction are the main concerns as discussed before, which became the main intention to design hip joint simulators [21]. In 2001 after a long debate between different experts in hip implant surgery, manufacturing, and testing, a new ISO standard was introduced to cover all concerns of wear testing and wear test simulator requirements for total hip replacements [37]. This newly introduced standard affected recent simulator designs.

Using coordinate measurement profilers Tuke and his colleagues measured the wear on a femur head [38]. Trommer and Maru [39] designed their eight-station tester, Zanini et al. [40] used X-ray to measure wear, Partridge et al. [41] used their setup called Pro-sim which has flexibility in both cup and femur holders. Again Viitala and Saikko [42] improved the HUT simulators.

## METHODOLOGY

### A. An introduction to Wear of total hip-joint prostheses standard (ISO 14242)

ISO 14242 giving general comments and some technical prescriptions about the wear of total hip-joint prostheses. This standard consists of four main parts[43]. In the first part, different loadings and displacement values for the test were presented. Also, the environmental condition of the test was maintained. In the second part, the measurement method is clarified. The third part of the standard is on a specific type of wear machine for a test, the so-called orbital bearing type. This is the most prevalent type of wear testing machine. The fourth part of the standard is the most recent document as well, introduce mispositioning of the femoral head. This standard is investigating such effects in wear.

The loadings and conditions in this paper investigated based on the values of the standard compared with literature.

### B. Total Hip Arthroplasty Design Exploration

Total hip replacement implants consist of two main components which are acetabular and femoral. Each component has its subcomponents as well. Components of acetabular are the acetabular shell and the liner, while components of the femoral are the neck, the femoral stem, and the femoral head. Acetabular shell, liner, femoral head, neck, and femoral stem, respectively, are connected to form THA.

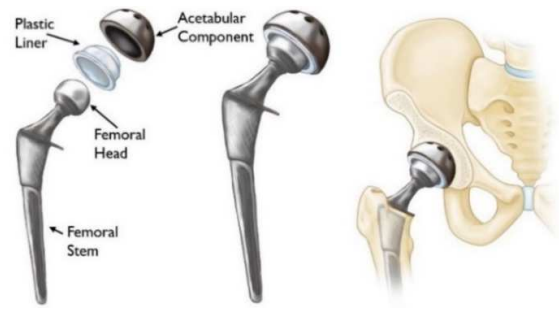


Fig. 1: Total Hip replacement components as implication [44]

A typical THA is shown in Fig. 1. These components are constrained by their geometry. By this means, stability is provided without limiting to necessary movements of the hip joint [44]. There are two main design variations for THA. Some of them are designed as a single part, called Monoblock, and some other design as a modular assembly. In the modular method, all components mentioned above are designed separately. These two methods have some advantages and disadvantages compared to each other. The modular method provides to design different geometrical relations between femoral components. Thanks to that, this method can be customized for patients due to wider geometric variations. There are two possible application methods during the surgery which are cemented and cementless during the THA operation. The methods lead to different geometrical shapes. For example, in the cementless operation, the acetabular cup has screw holes. These holes let acetabular cups have different mechanical properties and geometry.

Due to wear, induced by forces of friction, debris is formed. The amount of wear is a particularly important parameter [45]. The debris is not very well tolerated by the body and causes adverse effects, such as aseptic loosening of the joints and requiring revision surgeries [46]. For this reason, preclinical testing is mandated. This is mostly performed by man-made simulators, mimicking the forces and range of motion values, by the International Standards Organization ISO 14242 for assessing the amount of material loss due to the wear. In ISO 14242, gait cycle force and range of motion conditions are simulated. However, today subject-specific implants are being favored considering the variations in the activities. To create a realistic testing environment, it is especially important to provide realistic test conditions and include different scenarios to estimate the wear. This indicates, there is a potential for improving simulator designs, enabling them to accept different daily life activities as a test condition.

This article is a part of a research study that aims to understand the wear mechanisms of hip prostheses. ISO 14242 gives a good firm to gather information from an in vivo test. The final goal is to mimic real daily activity motions and designing a prosthesis based on data acquired from the simulator designed here.

### C. Measured hip load scenarios and comparison

ISO 14242-3 defined a load case based on the various measurements as mentioned in the previous part. Fig. 3 shows the load case for the test machine. This load case must be applied by a frequency of 1Hz. The resultant force for primary studies [17], [21], [47] compared with ISO 14242 [37].

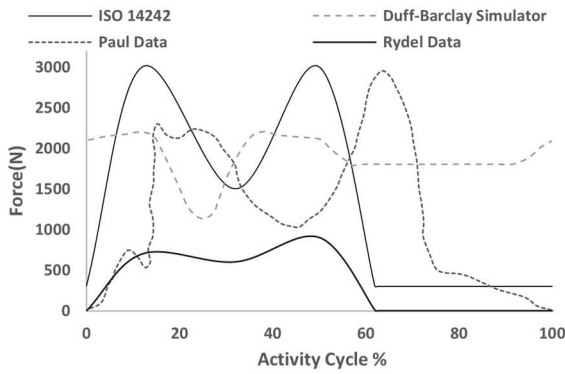


Fig. 3: REsultant force VS Gaitcycle percentage [17], [21], [37], [47]

Interestingly, the Duff Barclay simulator [21] has such a different simulated load regarding experimental data from Paul [47] and Rydell [17].

Fig. 2 shows some recent studies on motion capture methods that show interesting resultant forces that could be useful for further research [19].

#### D. Hip Simulator Architecture

According to the designs explored in this research, a schematic of the system was generated in Fig. 4. This schematic is a system architecture that helps understand how such a system behaves. An ideal hip simulator must rotate the femur head around its principal axis and give position feedback from the motion data. Also, it is necessary to have a closed-loop force control by getting resultant force and principal axis normal forces. Such a mechatronic system needs close interaction between the mechanical parts and the

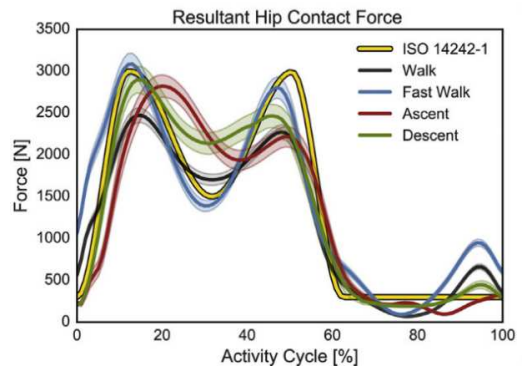


Fig. 2: Hip resultant forces measured by Motion capture [19]

electrical actuators, as well as the control system. In this study, the system is considered an electromechanical system. In some obsolete simulators, pneumatic actuators are also implemented which are not considered in this research.

#### E. Femur head load calculation

Although there are comprehensive research studies conducted on simulation of femur head forces, this section aims to get an understanding of the sensitivity of the simulator and the main factors affecting the design.

According to different studies, the femur head is considered as a hemisphere [48]–[52], therefore, a simplified model of a femur would be a hemisphere with a small eccentricity.

The schematic of the hip joint with applied force is depicted in Fig. 5.

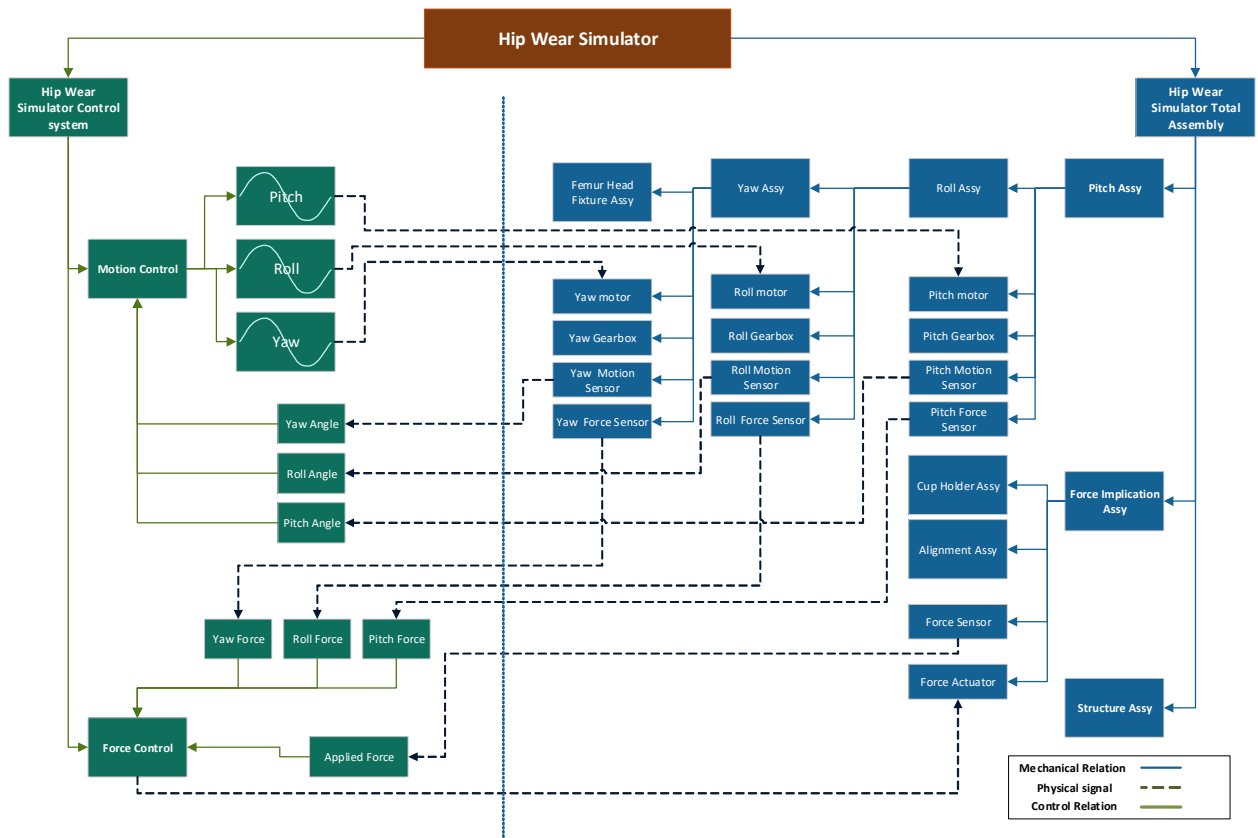


Fig. 4: Hip Wear Simulator system Architecture

TABLE I: REPORTED FRICTION COEFFICIENT FOR DIFFERENT MATERIAL PAIRS USING IN HIP REPLACEMENT [56]

Couple	Friction coefficient
316 L Stainless steel/UHMWPE	0.10 ± 0.02
Co-Cr alloy/UHMWPE	0.11 ± 0.02
A10 ceramic/UHMWPE	0.06 ± 0.01
Zirconia ceramic/UHMWPE	0.08 ± 0.02
Alumina ceramic/alumina ceramic	0.05 ± 0.01
Commercially available (Al <sub>2</sub> O <sub>3</sub> /6% ZrO <sub>2</sub> ) ceramic/commercially available (Al <sub>2</sub> O <sub>3</sub> /6% ZrO <sub>2</sub> ) ceramic	0.05 ± 0.01
A12 ceramic/A12 ceramic	0.05 ± 0.01
A10 ceramic/A10 ceramic	0.10 ± 0.02
Artificially aged A10 ceramic/artificially aged A10 ceramic	0.19 ± 0.04
Zirconia ceramic/zirconia ceramic	0.47 ± 0.06

The force  $F$  is acting force on the joint causes normal force  $N$  and frictional force  $f$  with source equivalent friction of  $\mu$ . Angle  $\theta$  is the angle of contact point compared to the acting force direction. The moment acting on the joint center would be

$$\sum F_x = -N \sin \theta_0 + \mu N \cos \theta_0 = 0 \quad (1)$$

$$\sum F_y = N \cos \theta_0 + \mu N \sin \theta_0 = F \quad (2)$$

$$\sum M_0 = \mu N R = F l \quad (3)$$

From (1):

$$\tan \theta_0 = \mu \quad (4)$$

Therefore, from (2), (3):

$$N = \frac{F}{\sqrt{1 + \mu^2}} \rightarrow \frac{l}{R} = \frac{\mu}{\sqrt{1 + \mu^2}} \quad (5)$$

For an acting moment on point  $O$ :

$$M = \mu N R = F R \frac{\mu}{\sqrt{1 + \mu^2}} \quad (6)$$

The equation (6) gives a rough approximation for moments acting in a hip prosthesis joint[53].

#### DESIGN EXPLORATION AND PARAMETER RESULTS

In this section, a brief analysis of different parameters affecting the system is explained. The parameters such as acting force  $F$ , the radius of femur  $R$ , and the friction of the two contact faces  $\mu$ .

According to the literature [54], the normal diameter of the femur of a grown person is between 28mm to 36mm. Femoral head diameters less than 28mm are considered as small femurs diameters above 36mm considered as a large femur head. The

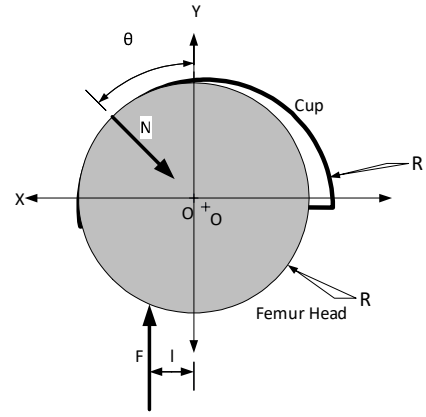


Fig. 5: 2D schematic of the hip joint with forces [53]

largest femur head diameter is reported as 54mm and the smallest one is about 22mm [55]. According to this, the maximum radius of the femur would be 27 mm and the minimum is 11 mm (radius is half of the diameter).

The friction coefficient of the joint is another key factor, according to the literature, the following values have been found for the varied materials and designs.

Based on the test requirement of ISO 14242, the normal force should be a maximum of 3000N. The test frequency of the simulator is 1Hz.

Using the values of Table I and Table II for friction coefficient, and other discussed parameters, by a Matlab code, sensitivity graphs were generated to understand the effects of each parameter better. Considering a friction coefficient from up to one and a resultant force up to 5kN, Fig. 6 generated.

Based on the calculated values, for the friction of 0.8, the maximum torque is 67 Nm.

However, there is another parameter that needs to be investigated, the frequency of the system, repeating the physical activity. Investigating power consumption gives an approximation of the sizes of the motors and the total size of the system.

According to the standard, the test frequency is about 1Hz, which is 6.28 rad/s. The power sensitivity to the frequency was also investigated. For a system with 60% efficiency (Fig. 7). This graph shows that this system is linearly sensitive to the frequency, and by increasing the frequency of the system, power consumption increases.

TABLE II: FRICTION RESULTS FROM THE STUDIES ON HIP REPLACEMENTS [50]

Prosthesis types	Friction results, $\mu$	Comments
McKee-Farrar	0.10-0.20	Static and dynamic loading
Charnley-Muller	0.05-0.20	Serum/synovial fluid lubrication
McKee Farrar	0.17	High $f$ for the explanted McKee-Farrar due to greater equatorial contact
McKee-Farrar	0.1-0.8	Dynamic loading, synovial fluid
Charnley	0.03-0.05	Static and dynamic loading
McKee-Farrar	0.25	Synovial fluid/silicone fluid
Charnley	0.05	Steady load
Charnley	0.053	Dynamic load, silicone fluid: $\eta = 0.02$ Pas
48 Charnley	0.040	High $f$ of the Kirchner was due to the degradation of the femoral head
Howmedica	0.02-0.03	Dynamic loading, water
Biomet	0.05-0.06	Dynamic loading, soln. of CMC
Kirchner	0.05-0.07	Static and dynamic loading
Charnley	0.12-0.15	Serum/synovial fluid lubrication
Charnley	0.05	High $f$ for the explanted McKee-Farrar due to greater equatorial contact

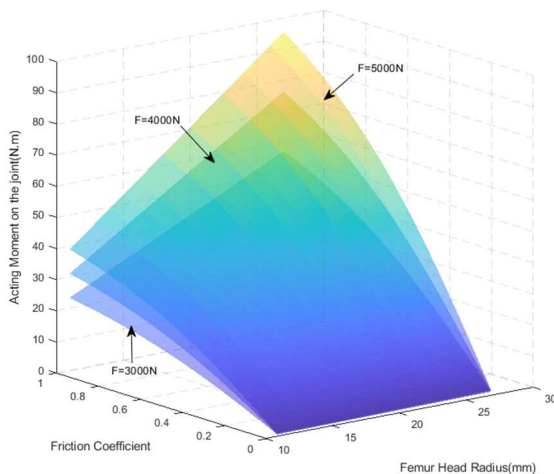


Fig. 6: Torque variation for different loads in different friction coefficient and femur head size

## DISCUSSIONS

The analyzed system architecture shows the relation between system control and mechanics. The wear simulator is an electromechanical system, thus, there is a close dependency between the system actuators in different degrees of freedom and the force control. Both force control and motion control of the system need a good synchronization as well since many primitive systems lacked in synchronization, therefore many problems were reported on those designs. The design reached maturity in the mechanical part but authors believe that system control has potential for more search and needs more explorations. Such explorations might lead to new designs in the mechanical side iteratively.

The force and power sensitivity analysis shows that by increasing the acting force on the femoral head, the acting moments increase drastically, specifically if there is a coefficient of friction larger than 0.8. This finding shows there is a need to consider the effect of the friction on the system motor selection to get the best system functionality. The femoral head size is amplifying the effects of surface friction as well. The largest femoral head radius is about 27mm it could be considered as a constant in design. The effect of test frequency on actuator selection is quite prominent. By increasing the frequency, the power consumption on each design point changed dramatically. Meaning that for instance by doubling the frequency, the system size becomes double. The frequency of the system is related to the speed of repetition of the simulated activity. High frequencies mimic fast activities such as running. Therefore, for such a system, it is important to know the intended use of the prosthesis. By increasing frequency in the wear test simulator, the system might test special prostheses intended for sports people.

## CONCLUSIONS

This research aims to conclude the design trend line of the hip simulators and the measurement of the hip joint loads. The general architecture of the system presented and the effects of the main factors on moments and power consumption of the system are also investigated to get a better idea about the design thresholds. This research is a road map for later research soon.

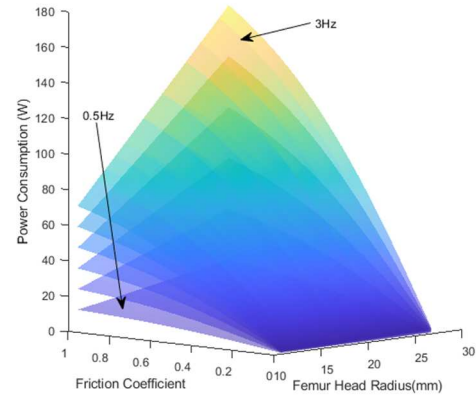


Fig. 7: Power sensitivity to Speed

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## REFERENCES

- [1] J. Chamley, *Low Friction Arthroplasty of the Hip*. Springer-Verlag Berlin Heidelberg, 1979.
- [2] G. Galilei, *Discorsi e dimostrazioni matematiche intorno a due nuove scienze*. 1638.
- [3] É.-J. Marey, *De la locomotion terrestre chez les bipèdes et les quadrupèdes*. Editeur inconnu, 1873.
- [4] G. Carlet, *Essai expérimental sur la locomotion humaine*. Impr. E. Martinet, 1872.
- [5] G. Demeny and É. A. V. A. Quénu, "Etude de la locomotion humaine dans les cas pathologiques," *Stm. Physiol.*, vol. 4, pp. 1887–89, 1888.
- [6] R. P. Schwartz and W. Vaeth, "A method for making graphic records of normal and pathologic gaits: Description of a new apparatus, the basograph," *J. Am. Med. Assoc.*, vol. 90, no. 2, pp. 86–89, 1928, doi: 10.1001/jama.1928.02690290016005.
- [7] R. P. Schwartz, A. L. Heath, and J. L. Wright, "ELECTROBASOGRAPHIC METHOD OF RECORDING GAIT," *Arch. Surg.*, vol. 27, no. 5, p. 926, 1933, doi: 10.1001/archsurg.1933.01170110111009.
- [8] R. Schwartz, A. Heath, W. Misiek, and J. Wright, "KINETICS OF HUMAN GAIT," *J. Bone Jt. Surgery, Am. Vol.*, vol. 16, pp. 343–350, 1934.
- [9] T. F. Besier, D. L. Sturmeiks, J. A. Alderson, and D. G. Lloyd, "Repeatability of gait data using a functional hip joint centre and a mean helical knee axis," *J. Biomech.*, vol. 36, no. 8, pp. 1159–1168, 2003, doi: 10.1016/S0021-9290(03)00087-3.
- [10] T. Y. Tsai, J. S. Li, S. Wang, D. Scarborough, and Y. M. Kwon, "In-vivo 6 degrees-of-freedom kinematics of metal-on-polyethylene total hip arthroplasty during gait," *J. Biomech.*, vol. 47, no. 7, pp. 1572–1576, 2014, doi: 10.1016/j.jbiomech.2014.03.012.
- [11] W. Platzer, *Color Atlas of Human Anatomy (Vol 1. Locomotor System) 7th Edition*. Thieme, 2014.
- [12] J. F. Stoltz, *Regenerative Medicine and Cell Therapy*. 2012.
- [13] H. Meyer, "Die Architectur der Spongiosa Reichert und Dubois Reymonds," *Virchow's Arch.*, p. 615, 1867.
- [14] A. Fick, "Statische Betrachtung der Muskulatur des Oberschenkels," *Gesammelte Schriften*, vol. 1, pp. 415–424, 1904.
- [15] H. MILCH, "PHOTO-ELASTIC STUDIES OF BONE FORMS," *J. Bone Jt. Surg.*, vol. 22, no. 3, p. 621, 1940.
- [16] H. Debrunner, "Ueber einige orthopädische Probleme," *Praxis (Bern 1994)*, vol. 47, pp. 262–266, 1958.
- [17] N. W. Rydell, "Forces acting on the femoral head-prosthesis. A study on strain gauge supplied prostheses in living persons," *Acta Orthop. Scand.*, vol. 37, 1966, doi: 10.3109/ort.1966.37.suppl-

- 88.01.
- [18] B. E. Ingelmark and E. Blomgren, "An apparatus for the measurement of pressure, especially in human joints," *Uppsala Lakareforen Forh.*, vol. 53, no. 1–2, pp. 75–94, 1948.
- [19] D. E. Lunn, E. De Pieri, G. J. Chapman, M. E. Lund, A. C. Redmond, and S. J. Ferguson, "Current Preclinical Testing of New Hip Arthroplasty Technologies Does Not Reflect Real-World Loadings: Capturing Patient-Specific and Activity-Related Variation in Hip Contact Forces," *J. Arthroplasty*, vol. 35, no. 3, pp. 877–885, 2020, doi: 10.1016/j.arth.2019.10.006.
- [20] B. W. Stansfield, A. C. Nicol, J. P. Paul, I. G. Kelly, F. Graichen, and G. Bergmann, "Direct comparison of calculated hip joint contact forces with those measured using instrumented implants. An evaluation of a three-dimensional mathematical model of the lower limb," *J. Biomech.*, vol. 36, no. 7, pp. 929–936, 2003, doi: 10.1016/S0021-9290(03)00072-1.
- [21] B. I. Duff-barclay and D. T. Spillman, "Total human hip joint prostheses- a laboratory study of friction and wear," vol. 181, no. 3, 1966.
- [22] D. Dowson, P. S. Walker, M. D. Longfield, and V. Wright, "A joint simulating machine for load-bearing joints," *Med. Biol. Eng.*, vol. 8, no. 1, pp. 37–43, 1970, doi: 10.1007/BF02551747.
- [23] P. S. Walker and B. L. Gold, "The tribology (friction, lubrication and wear) of all-metal artificial hip joints," *Clin. Orthop. Relat. Res.*, vol. 17, pp. 285–299, 1971, doi: 10.1097/00003086-199608001-00002.
- [24] J. H. Dumbleton, D. A. Miller, and E. H. Miller, "A simulator for load bearing joints," *Wear*, vol. 20, no. 2, pp. 165–174, 1972, doi: 10.1016/0043-1648(72)90379-1.
- [25] B. Weightman, S. Simon, I. Paul, R. Rose, and E. Radin, "Lubrication Mechanism of Hip Joint Replacement Prostheses.," *ASME Pap 72-Lub-D meet 1972*, pp. 131–135, 1972.
- [26] S. A. V Swanson, M. A. R. FREEMAN, and J. C. Heath, "LABORATORY TESTS ON TOTAL JOINT REPLACEMENT PROSTHESES," *J. Bone Jt. Surg.*, vol. 55B, no. 4, pp. 759–773, 1973.
- [27] M. Ungethüm, J. Hildebrandt, M. Jäger, and H. G. Moslé, "Ein neuer Simulator zur Testung von Totalendoprothesen für das Hüftgelenk," *Arch. Orthop. Unfallchir.*, vol. 77, no. 4, pp. 304–314, 1973, doi: 10.1007/BF00418916.
- [28] G. Beutler, M. Lehmann, and G. Stahlh, "Wear Behaviour of Medical Engineering Materials," *Wear*, vol. 33, pp. 337–350, 1974.
- [29] M. A. Swikert and R. L. Johnson, "SIMULATED STUDIES OF WEAR AND FRICTION IN TOTAL HIP PROSTHESIS COMPONENTS WITH VARIOUS BALL SIZES AND SURFACE FINISHES," Cleveland, Ohio, 1976.
- [30] A. Cappozzo, L. Cini, A. Pizzoferrato, C. Trentani, and S. S. Cortesi, "Evaluation of hip arthroprostheses by means of body environment simulators," *J. Biomed. Mater. Res.*, vol. 11, no. 5, pp. 657–669, 1977, doi: 10.1002/jbm.820110503.
- [31] H. A. McKellop and I. C. Clarke, "Evolution and evaluation of materials-screening machines and joint simulators in predicting in vivo wear phenomena," *Funct. Behav. Orthop. Biomater.*, vol. 2, pp. 51–85, 1984.
- [32] D. Dowson and B. Jobbins, "Design and development of a versatile hip joint simulator and a preliminary assessment of wear and creep in Charnley total replacement hip joints," *Eng. Med.*, vol. 17, no. 3, pp. 111–117, 1988, doi: 10.5040/9781474221467.part-002.
- [33] V. Saikko, P. Paavolainen, M. Kleimola, and P. Slätis, "A Five-Station Hip Joint Simulator for Wear Rate Studies," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 206, no. 4, pp. 195–200, 1992, doi: 10.1243/PIME\_PROC\_1992\_206\_291\_02.
- [34] V. Saikko, "A Simulator Study of Friction in Total Replacement Hip Joints," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 206, no. 4, pp. 201–211, 1992, doi: 10.1243/PIME\_PROC\_1992\_206\_292\_02.
- [35] V. O. Saikko, "A three-axis hip joint simulator for wear and friction studies on total hip prostheses," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 210, no. 3, pp. 175–185, 1996, doi: 10.1243/PIME\_PROC\_1996\_210\_410\_02.
- [36] V. Saikko, "A 12-station anatomic hip joint simulator," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 219, no. 6, pp. 437–448, 2005, doi: 10.1243/095441105X34419.
- [37] C. Kaddick and M. A. Wimmer, "Hip simulator wear testing according to the newly introduced standard ISO 14242," *J. Biomech.*, vol. 34, no. SUPPL. 1, pp. 429–442, 2001, doi: 10.1016/s0021-9290(04)00168-x.
- [38] M. Tuke, A. Taylor, A. Roques, and C. Maul, "3D linear and volumetric wear measurement on artificial hip joints - Validation of a new methodology," *Precis. Eng.*, vol. 34, no. 4, pp. 777–783, 2010, doi: 10.1016/j.precisioneng.2010.06.001.
- [39] R. M. Trommer and M. M. Maru, "Importance of preclinical evaluation of wear in hip implant designs using simulator machines," *Rev. Bras. Ortop. (English Ed.)*, vol. 52, no. 3, pp. 251–259, 2017, doi: 10.1016/j.rboe.2016.07.004.
- [40] F. Zanini, S. Carmignato, E. Savio, and S. Affatato, "Uncertainty determination for X-ray computed tomography wear assessment of polyethylene hip joint prostheses," *Precis. Eng.*, vol. 52, no. February, pp. 477–483, 2018, doi: 10.1016/j.precisioneng.2018.02.009.
- [41] S. Partridge, J. L. Tipper, M. Al-Hajjar, G. H. Isaac, J. Fisher, and S. Williams, "Evaluation of a new methodology to simulate damage and wear of polyethylene hip replacements subjected to edge loading in hip simulator testing," *J. Biomed. Mater. Res. - Part B Appl. Biomater.*, vol. 106, no. 4, pp. 1456–1462, 2018, doi: 10.1002/jbm.b.33951.
- [42] R. Viitala and V. Saikko, "Effect of random variation of input and various daily activities on wear in a hip joint simulator," *J. Biomech.*, vol. 106, pp. 1–7, 2020, doi: 10.1016/j.jbiomech.2020.109831.
- [43] "ISO 14242 Implants for surgery — Wear of total hip-joint prostheses."
- [44] G. Jung, Y. Kim, N. T. Khan, and Y. B. Kim, "Risk evaluation method for range of motion after total hip arthroplasty applying monte-carlo simulation," *Procedia Comput. Sci.*, vol. 141, pp. 351–357, 2018, doi: 10.1016/j.procs.2018.10.190.
- [45] S. Braun *et al.*, "Backside wear in acetabular hip joint replacement," *Acta Biomater.*, vol. 83, pp. 467–476, 2019, doi: 10.1016/j.actbio.2018.10.045.
- [46] E. D. Pawar, A. K. Yadav, A. Sharma, and A. Harsoor, "Ceramic liner fracture in ceramic on ceramic Total hip arthroplasty: A case report," *Int. J. Surg. Case Rep.*, vol. 65, pp. 242–244, 2019, doi: 10.1016/j.ijscr.2019.10.079.
- [47] J. P. Paul, "Force actions transmitted by joints in the human body," in *Proceedings of the Institution of Mechanical Engineers, Conference Proceedings*, 1966, vol. 181, no. 33, pp. 8–15, doi: 10.1243/PIME\_CONF\_1966\_181\_201\_02.
- [48] M. Ali and K. Mao, "Contact mechanics and wear simulations of hip resurfacing devices using computational methods," *Acta Bioeng. Biomech.*, vol. 16, no. 2, pp. 103–110, 2014, doi: 10.5277/abb140212.
- [49] F. Liu, J. Fisher, and Z. Jin, "Effect of motion inputs on the wear prediction of artificial hip joints," *Tribol. Int.*, vol. 63, pp. 105–114, 2013, doi: 10.1016/j.triboint.2012.05.029.
- [50] R. M. Hall and A. Unsworth, "Friction in hip prostheses," *Biomaterials*, vol. 18, no. 15, pp. 1017–1026, 1997, doi: 10.1016/S0142-9612(97)00034-3.
- [51] T. Roy, D. Choudhury, S. Ghosh, A. Bin Mamat, and B. Pinguan-Murphy, "Improved friction and wear performance of micro dimpled ceramic-on-ceramic interface for hip joint arthroplasty," *Ceram. Int.*, vol. 41, no. 1, pp. 681–690, 2015, doi: 10.1016/j.ceramint.2014.08.123.
- [52] V. Pakhaliuk, A. Polyakov, M. Kalinin, and V. Kramar, "Improving the finite element simulation of wear of total hip prosthesis' spherical joint with the polymeric component," *Procedia Eng.*, vol. 100, no. January, pp. 539–548, 2015, doi: 10.1016/j.proeng.2015.01.401.
- [53] A. Faraz and S. Payandeh, "General Friction Models of Joints," in *ENGINEERING APPROACHES FOR MIS*, 2000.
- [54] S. Affatato, "Contemporary designs in total hip arthroplasty (THA)," in *Perspectives in Total Hip Arthroplasty*, Elsevier, 2014, pp. 46–64.
- [55] J. Girard, "Femoral head diameter considerations for primary total hip arthroplasty," *Orthopaedics and Traumatology: Surgery and Research*, vol. 101, no. 1. Elsevier Masson SAS, pp. S25–S29, Feb. 01, 2015, doi: 10.1016/j.otsr.2014.07.026.
- [56] V. Banchet, V. Fridrici, J. C. Abry, and P. Kapsa, "Wear and friction characterization of materials for hip prosthesis," *Wear*, vol. 263, no. 7-12 SPEC. ISS., pp. 1066–1071, 2007, doi: 10.1016/j.wear.2007.01.085.