# BIOMECHANICAL ANALYSIS OF HIP AND KNEE JOINTS DURING GAIT IN ELDERLY SUBJECTS

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## SUMMARY

The objective of this study was to quantify the range of motion, force momentum, power and the mechanical work performed by hip and knee joints during gait in a group of subjects aged between 55 and 75 years. As a common activity of daily life, walking is often prescribed as a therapeutic exercise in elderly adults' rehabilitation. Kinematic and kinetic analyses during gait were obtained from optical tracking, force plate, standardized x-ray imaging and anthropometric data. The total effort generated by the hip joint during gait was greater than the one of the knee joint. The hip joint generated a total effort of 0.40J/kg, with 22% on the frontal plane, 76% on sagittal plane, and 2% on transverse plane. The total effort generated at the knee joint during gait was 0,30J/kg, with 7% occurring on frontal plane, 90% on sagittal plane, and 3% on transverse plane. The biomechanical analysis of the joints during different activities would help clinicians to identify and understand important variables required for improving the performance and deficits of elderly individuals.

***Keywords:*** *Biomechanics, Gait, Kinetics, Kinematics, Elderly*

## INTRODUCTION

Elderly population's growth in Brazil is evident. It is estimated that, by 2025, Brazil will be the sixth country in the world with the highest number of elderly individuals. Aging leads to an increased incidence of chronic diseases, which, if not treated or properly monitored, may leave permanent sequels, disabling the elderly and making them lose their autonomy and functional independence[1]. Gait changes are frequent problems as we age[2]. Improving or even maintaining a functional gait is a challenging task and of great concern for healthcare professionals.

While walking is an unconscious movement and almost automatic, it is also very complex, because it requires body to be in perfect harmony inside, offsetting external forces that are continuously acting on our segments[3]. For biomechanical analysis, a description and measurements of the strengths producing motion are required. The areas of mechanics describing gait movements are temporal, spatial and kinematic data. Kinematic variables include acceleration, speed and angle shift. The angle shift describes joint ranges of motion during gait[4].

The study of the forces acting on our body is called kinetics[4]. Kinetic variables include ground response forces, force momentum, power and the mechanical work performed by joints during gait[5]. Force momentum characterizes the sum of muscles, tendons, ligaments and bones strengths internally acting to oppose external forces acting on our body. Therefore, the magnitude of internal force momentum reflects how likely muscle and passive strengths would cause a segment to rotate over its rotation center[5]. Power is a product of the force momentum X angle speed, and it is the only kinetic variable expressing muscular function when a muscle is tensioned, that is, if they are concentrically or eccentrically contracting during a given function[6]. To measure the mechanical work, the power product times the elapsed time is required. If the work is positive, it means that a muscle generated energy into the system by means of a concentric contraction. If it is negative, it means that the muscle absorbed energy from system by means of an eccentric contraction. That energy flow or exchange is the amount of mechanical work performed by each joint during a given task[4].

The most common method for kinetic data analysis is the inverse dynamics which, by Newton's equations, applies the variables known as ground response strength, segments' mass and mass center, inertia momentum and position of segments on the free body diagram[3]. That analysis allows for the calculation of the sum of force momentum, power and mechanical work of joints from distal to proximal, that is, from ankle to knee and hip.

Reduced speed and step extension, ranges of motion and force momentum in hip and knee joints that occur in gait with age have already been described[2;7]. However, few studies describe hip and knee joints power and mechanical work during gait, and, particularly, few studies emphasize the biomechanical analysis on the three motion planes [6]. Thus, the objective of this study was to describe kinematic and kinetic characteristics of hip and knee joints in a group of healthy subjects aged 55-75 years on the three motion planes. Biomechanical mechanisms involved in human gait must be explained in order to understand the consequences of failures or absences of these in aging process, thus rendering us able to take more effective therapeutic measures, enabling the elderly to have a better quality of life.

## MATERIALS AND METHODS

### Subjects

Data concerning 30 healthy subjects, with no previous history of gait pathologies or changes, aged 55 to 75 years were collected. The inclusion criteria included: individuals above the age of 55, with no previous history of osteoarthritis or knee and hip pain, who could ambulate without any mechanical aid. The group of subjects was constituted of community members, who were recruited by responding to an advertisement published on local newspapers.

### Measurement System

The *Queen's Gait Analysis in Three Dimensions* (QGAIT – Queen's University - Kingston, Ontario, Canada) system was employed for collecting kinematic and kinetic data and for measuring power and mechanical work data during gait. QGAIT system's methodology, validity, and reliability have already been published in details[8]. In order to find the exact location of hip and knee rotation center, the QUESTOR *Precision Radiographs* (QPR-Queen's University. Kingston, Ontario, Canada) X-ray was employed in the present study. The validity and reliability of it have been previously explained[9]. Infrared headlights were placed over bone protuberances of the major trochanter, femoral lateral epicondyle, fibular head, and lateral maleollus. Two projectiles, each one built with a headlight on the tip, were placed fixed on thigh and leg segment by means of an elastic band covered with Velcro. Three marks for each segment are required to design the direction in three dimensions of thigh and leg segments in space. A 16-channel analogical/ digital converser made the integration of *Optotrak* (Northern Digital Inc., Waterloo, Ontario, Canada) system, the force platform (AMTI, Advanced Mechanical Technology, Inc.), the footswitch and the anthropometric data (segments' mass, mass center and inertial properties) for measuring the force momentum as Newton-meters (Nm) during gait.

Hip and knee force momentum was collected by using inverse dynamics[3]. In inverse dynamics, segments are regarded as stiff bodies and joints as hinges. In our model, the foot was regarded as a portion of the leg, thus, our model was constituted of leg and thigh segments[8]. Force momentum was first measured on the knee and then on the hip. Force momentum signs were determined according to the right hand rule, where the thumb follows the direction of the axis indicated at the top of graphs and the hand spins around the axis. For example, at sagittal plane, by positioning the right thumb on joint axis from lateral to medial (LM), fingers rotate positively, in the same direction as knee flexion. Therefore, positive is an external force momentum that tends to flex knee, which is balanced by an extensor internal momentum of similar magnitude but opposite enough to balance the system. Now, on the hip, a positive external momentum tends to extend the hip; therefore, a flexor internal momentum would be required to balance external forces.

Power (watts) was measured as the product of force momentum (Newton-meters) and angle speed (radians/sec), and work (Joule) was measured by means of the integration of power over time[4]. Variables power and work were normalized by body weight, resulting in unit measurements W/kg and J/kg, respectively. Total mechanical work of each joint during gait cycle was measured by summating absolute values of positive and negative work made during gait.

For angles measurement, a floating axis system described by Grood et al.[10] was employed. Abduction, flexion and internal rotation were regarded as positive angles both for hip and knee.

The present study was approved by Queen's University Committee of Ethics (Ontario, Canada), and all subjects signed a free and informed consent term before admission. Subjects were first sent to the X-ray service, where three anatomical marks (major trochanter, femoral lateral epicondyle, and fibular head) were identified and highlighted with a washable tint pen, and a small plumb ball was placed over these protuberances with the aid of a tape. At the motion analysis laboratory, the headlights were placed just on the marks made with the pen and previously identified on X-ray, in addition to the marks previously described. Subjects were asked to ambulate with comfortable shoes, and a footswitch was placed under the right heel sole in order to determine contact and contact loss of feet to the ground. Subjects were recommended to walk normally on the runway, always keeping the right arm on the chest to avoid major trochanter mark fading. The step of interest started with the right foot touching the force platform and ended when that very right foot touched the ground again.

### Statistical analysis

Time and anthropometric data between female and male groups were compared by using the t-test with a significance level of $p > 0.05$. Kinematic and kinetic data for the whole group were descriptively compared with similar studies found in literature.

### RESULTS

Data concerning 17 male patients and 13 female subjects were assessed. The mean age for the female group was $63.8 \pm 6.3$ years and for the male group, $66.6 \pm 5.6$ years ($p > 0.27$). The mean age for the 30 subjects was $65.1 \pm 6.0$ years (range: 55 - 75 years). The average step speed for female group was $1.13 \pm 0.12$ m/sec and, for male group, $1.35 \pm 0.18$ m/sec ($p > 0.36$). The total speed average for the group as a whole was $1.17 \pm 0.18$ m/sec. Figures 1, 2 and 3 show a graphic representation of kinematic, kinetic and power data, respectively, and Table 1 shows power peaks and its association on gait cycle. The vertical bar indicates the average point in percentage of the gait cycle (62%) when fingers lost contact with the ground, i.e., toe-off.

At frontal plane, hip joint initiates the cycle in abduction, reducing range of motion (rom) soon after the heel hits the ground, and back to abduction at the end of pre-oscillation phase, reaching to a maximum of 10.4° at 71% of the cycle. Hip rom at frontal plane was 11.1°. The internal force momentum of the hip is adbuctive for balancing external forces that tend to adduct the hip. Two peaks were identified, the highest being 0.71Nm/kg and occurring at 47% of gait cycle. The hip shows four power peaks, with Q2F and Q4F being the most expressive ones in terms of magnitude, reaching 0.40 W/kg at 13% and 0.29W/kg at 54%, respectively.

Knee rom at frontal plane was 4.9°, with the highest abduction peak occurring at 83% of the cycle, and reaching 6.0°. Two peaks of internal abduction momentum are present in knee, with the maximum occurring at 17% of the cycle, reaching 0.38Nm/kg. Knee power at frontal plane from 0 to 15% of the cycle (J1F) characterizes a concentric control of abductors (0.07W/kg at 11%), changing to an eccentric control between 15 and 30% at the medium support (J2F -0.05W/kg at 21%) and between 40 and 50% at pre-oscillation phase (J3F -0.04 W/kg).

At sagittal plane, hip rom was 38.9°. The maximum extension occurred at pre-oscillation phase, reaching -16.2° at 53% and the maximum flexion was 22.7° when heel hit the ground. The internal extensor momentum to balance external flexor forces reached -0.76Nm/kg at 6% of the cycle, followed by an internal flexor momentum peak at 51%, reaching 0.64Nm/kg, once external forces trend is to extend hip. Hip H1S power peak corresponds to that of the extensors, acting concentrically from the phase of heel hit to end support phase. From 35% of the cycle, hip extensors start to act eccentrically (H2S), because external forces are now posterior to the hip. The flexors' concentric activity preparing the hip for oscillation phase, reaches a maximum peak of 0.62W/kg (H3S) at 62% of the cycle (Figure 3 and Table 1).

At sagittal plane, knee joint shows a rom of 60.4°. The maximum flexion occurred at 73% of the cycle, reaching 59.5° as early as the medium oscillation phase, and the maximum extension was -0.8° at 1% of the cycle. The highest force momentum peak balances external forces that tend to flex the knee, creating an internal extensor momentum of 0.20Nm/kg at 17% of the cycle. Two negative peaks occur at 3% and 42% of the gait cycle, with magnitudes of -0.34Nm/kg and -0.35Nm/kg, respectively, corresponding to the external extension forces being offset by internal flexion forces of the knee. Knee power at sagittal plane is marked by many peaks, starting with J1S from 0 to 10% of the cycle, where flexors assure a flexed knee soon after the heel hits the ground in order to help on absorbing impact. Then, extensors control becomes eccentric (J2S) followed by a concentric control (J3S) to assure the second lock of the knee in extension. J4S are flexors absorbing energy and breaking extensors, followed by J5S, where flexors are already acting concentrically to start oscillation. The first negative peak, J6S, correspond to extensor muscles acting eccentrically controlling the excessive knee flexion and, then, also eccentrically controlling the flexors assuring that the knee will touch the ground when extended (Table 1).

At cross-sectional plane, total rom of the hip was 10.5° and of the knee 12.9°, and the internal force momentum were predominantly external rotation in both joints. Hip power Q1T corresponds to internal rotators acting concentrically between heel hit and medium support, followed by Q2T where external rotator muscles of the hip are now acting eccentrically, slowing down hip internal rotation, preparing the hip for oscillation phase. At the knee, external rotators act concentrically at end support phase to unlock the knee and enable flexion during oscillation phase. Following, J2T occurs between 50 and 60% of the cycle at pre-oscillation phase, acting eccentrically to slow down external rotators.

The total work done by hip joint was 0.40 J/kg, where 22% occurred at frontal plane, 76% at sagittal plane, and 2% at cross-sectional plane. Knee joint generated a total work of 0.30 J/kg. Most of the work, 90%, was done at sagittal plane, followed by 7% at frontal plane and 3% at cross-sectional plane (Table 1).

## DISCUSSION

The objective of the present study was to describe gait features in a group of subjects aged from 55 to 75 years, in order to understand biomechanical changes that occur as we age. Although the group included a higher number of men compared to women, anthropometric differences (age) and time differences (gait speed) between both genders were not significant.

The reduced gait speed among the thirty subjects is consistent with studies using samples of similar age[2]. This reduction leads to range of motion loss in all joints, especially at sagittal plane[11]. In the present study, we see a reduction both on hip flexion when the heel hits the ground and on extension, at end support phase. These findings are consistent with studies that made kinematic comparisons between elderly and youngster groups during gait[11,12]. According to Kerrigan et al.[11], extension peak loss in the elderly remains, even with speed increase, and old individuals usually offset it by bending pelvis anteriorly. Anterior pelvic bending further reduces hip extension and increases trunk flexion, which threatens stability and precludes body to progress ahead[4]. Regarding knee joint at sagittal plane, a reduced flexion was noticed during oscillation, consequently leading to total rom loss during gait cycle. These findings are consistent with studies described in literature[13].



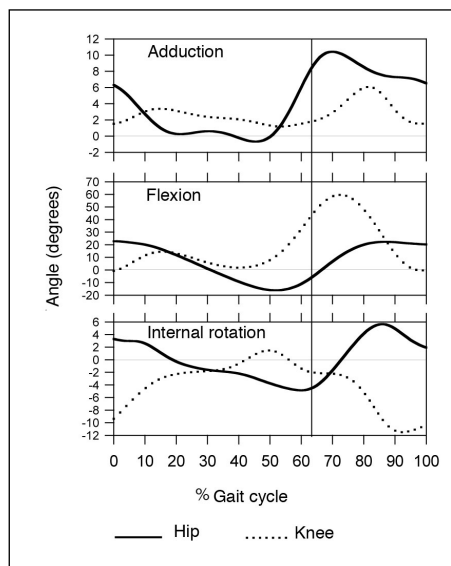**Figure 1-** *Angle shift (degrees) of hip and knee joints during gait (N = 30). A vertical line indicates toe-off.*
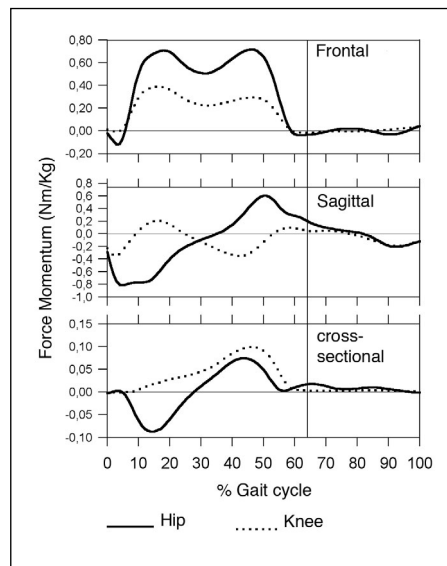


**Figure 2 -** *Force momentum (Nm/kg) of hip and knee joints during gait (N = 30). A vertical line indicates toe-off.*
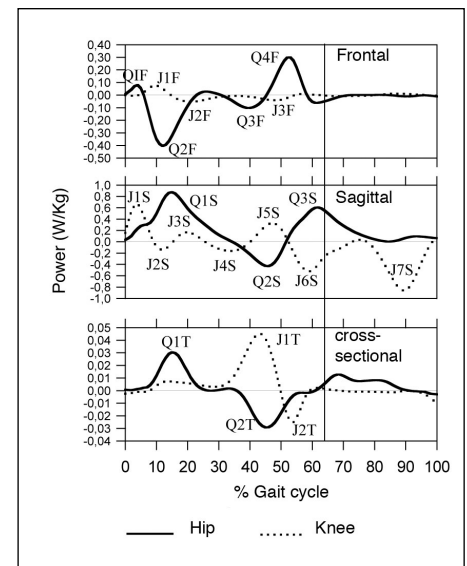


**Figure 3 -** *Power (W/kg) of hip and knee joints during gait (N = 30). A vertical line indicates toe-off.*

| Joint | Frontal Plane | | Sagittal Plane | | Cross-sectional plane | |
|---|---|---|---|---|---|---|
| Hip | Power (W/kg) | % | Power (W/kg) | % | Power (W/kg) | % |
| | Q1F=+0.08 | 5 | Q1S=+0.87 | 15 | Q1T=+0,03 | 15 |
| | Q2F=-0.40 | 13 | Q2S=-0.43 | 46 | Q2T=-0,03 | 45 |
| | Q3F=-0.10 | 41 | Q3S=+0.61 | 62 | | |
| | Q4F=+0.29 | 54 | | | | |
| Total work % | 21.6J/kg 22% | | 76.1J/kg 76% | | 2.3J/kg 2% | |
| Knee | J1S=+0.07 | 11 | J1S=+0.67 | 4 | J1T=+0,05 | 43 |
| | J2S=-0.05 | 21 | J2S=--0.14 | 12 | J2T=-0,03 | 54 |
| | J3S=-0.04 | 50 | J3S=+0.16 | 20 | | |
| | | | J4S=-0.16 | 34 | | |
| | | | J5S=+0.32 | 48 | | |
| | | | J6S=-0.53 | 59 | | |
| | | | J7S=-0.86 | 90 | | |
| Total work % | 6.6J/kg 7% | | 90.5J/kg 90% | | 2.9J/kg 3% | |

Q: hip; J: knee
F: frontal; S: sagittal; T: cross-sectional

**Table 1** - *Hip and knee power at frontal, sagittal and cross-sectional planes with the corresponding % at gait cycle ( n = 30).*

Judge et al.[14] measured hip rom at frontal and cross-sectional planes in a group of elderly individuals with mean age of 79 years. The findings showed higher adduction peaks, as well as a longer adductor phase on the hip during support. Regarding cross-sectional plane, the hip remained in internal rotation throughout the cycle[14], oppositely to our study. The differences reported may be associated to subjects' ages, once the mean age in our study was lower compared to the study by Judge et al.[14]. Another difference may be in determining joint rotation centers. Although in our study and in the study by Judge et al.[14] the same kind of system and fixed coordinate were used to determine joints angles, the way to determine rotation center was different. In the present study, the QPR[9] was employed, which accurately detects the position of hip and knee rotation center. But, in the study by Judge et al.[14], external marks were used as joint rotation centers. Errors in determining rotation centers may lead to discrepancies of joint measurements and of the force momentum for hip and knee joints[15;16].

Force momentum on hip and knee joints obtained in the current study generated curve patterns that are similar to studies described in literature[6;17]. The magnitude of force momentum of the hip at frontal plane reinforces the importance of abductors at the support phase of gait, by means of an eccentric control (Q2F) balancing medial external forces to the hip, and a concentric control (Q4F) supporting the opposite pelvis weight, precluding an excessive drop of gravity center. The total work generated at frontal plane by hip was relatively expressive (22%), reinforcing the importance of analysis at frontal plane, since it cam provide valuable information concerning muscle status of patients with hip pathologies.

The work done by hip at sagittal plane was greater (76%) when compared to other planes. The internal extensor momentum, at early support phase, generates energy due to extensors' concentric contraction, followed by an eccentric contraction of flexors breaking hip extension to enable the flexor phase of oscillation. Both the magnitude of the force momentum and hip power in the present study are lower when compared to magnitudes reported on the study by Eng et al.[6], which used a sample of young subjects.

Knee shows an internal abductor momentum representing ligaments' passive forces and fascia lata and iliotibial band tensor muscles' active forces, opposing to external adductor forces acting by adducting the knee at the support phase of gait[6]. First, a concentric control occurs, followed by two small peaks of energy absorption. Regarding the young individuals group data reported by Eng et al.[6], despite of similarities on curve pattern at force momentum between both studies, the young individuals group showed more expressive magnitudes. Regarding power analysis, that group on the study by Eng et al.[6] was characterized by only two peaks, when compared to the three peaks on the present study. Furthermore, total work at that plane was 11% compared to 7% in our study. It is possible that this extra peak of energy absorption generated by the elderly individuals is a strategy for providing a better control over external adductor forces during the support phase of gait. At the same time, we believe that the reduced hip joint work is probably related to weaknesses of abductor muscles in the elderly.

Similarly, knee force momentum at sagittal plane has also presented smaller magnitudes when visually compared to other study's results[6]. However, 90% of the work done by knee during the support phase of gait occurred at sagittal plane, 5% more when compared to the study by Eng et al.[6] using a young sample. The internal knee momentum when touching the ground is of flexion, which helps on absorbing impact, followed by an internal extensor momentum, when eccentric and concentric contractions oscillate, controlling knee semi-flexion and preventing an excessive vertical translation of the gravity center[18]. The next internal momentum is flexor, and occurs due to an increased gastrocnemius muscle activity on the ankle, acting eccentrically (J4S) and restraining tibial progression ahead the foot. With ankle lift, the internal flexor momentum is reduced, which starts to generate energy by concentrically contracting to remove the foot from the ground. When oscillation phase begins, while knee internal extensor force momentum is minimal in magnitude, the power generated shows expressive peaks, probably because of the angle speed involved in the phase of motion (Table 1). The first power peak (J6S) is the extensor, eccentrically slowing down leg's rotation, minimizing ankle lifting and hip flexion. At end oscillation, the action is eccentrically reverted to flexors (J7S) to assure an extended knee at heel impact[19].

The role of flexor and extensor muscles was largely emphasized by Sadeghi et al.[17] in a comparative study between youngsters and elderly subjects during gait. The authors showed that the key role of knee's flexor/ extensor muscles in elderly subjects was to control balance during simple support

phase, and, in the youngster subjects, its role was controlling both balance and propulsion.

In the present study, the results of kinetic analysis of the knee at frontal and sagittal planes, when observationally compared to the results found in the studies by Eng et al.[6] and Sadeghi et al.[17], show that muscular actions during gait in elderly individuals are functionally different. That is, elderly individuals tend to concentrate muscle functions aiming to maintain balance, but, since youngsters have an increased muscle control, they can distribute functions both to balance and to propulsion. In the present study, it seems that the strategy employed by elderly subjects was to reduce knee work at frontal plane and to increase it at sagittal plane, in an attempt to use a larger portion of knee flexor/ extensor muscles and keep balance during gait.

Although the magnitude of force momentum, power and work at cross-sectional plane are less expressive when compared to other motion planes, the analysis cannot be neglected, since it may clarify some important pathologic aspect. Hip data are consistent to the study by Winter et al.[20], where the initial internal momentum of internal hip rotation followed by an internal momentum of external rotation act as a slowing and speeding mechanism of the pelvis, respectively. The internal force momentum at knee cross-sectional plane in the present study was characterized by an external rotation momentum throughout support phase, which differs from the study by Eng et al.[6], but is consistent to the reports by Allard et al.[21], although the study group was constituted of young healthy individuals. Knee force momentum at cross-sectional plane is a response of knee ligaments restraining cross-sectional rotations of the hip and pelvis during the support phase of gait[6;20].

Some important points in the present study should be clarified. First, the QGAIT system regards foot as part of the leg segment; therefore, analysis of the ankle was not performed. According to data by Eng et al.[6], the work done by ankles during gait exceeds knee joint. Another important and limiting factor of the present study is concerned to the analysis of inverse dynamics, which ignores the co-contraction of antagonist muscles. In this case, both force momentum and power and work are measured using underestimated values. According to Andriacchi et al.[22], co-contraction is required to balance other external force momenta that are being applied on the segment during motion, thus generating better joint stability.

## CONCLUSION

The present study described the kinematics and kinetics of hip and knee joints in three dimensions during gait performed by 55-75 year-old individuals. While we did not intend to compare our data with literature reports on young individuals, some differences such as hip's and knee's range of motion loss and magnitude reduction on force momentum are clear. Furthermore, power analysis allows us to visualize the role of each joint during gait. The hip performs a stronger work at sagittal plane, followed by a relatively strong work at frontal plane, while the knee joint works almost only at sagittal plane. In terms of absolute values, the hip works stronger than knee during gait. It would be interesting that future studies perform biomechanical analysis on other activities for us to better understand the role of each joint during motion, thus finding further grounds to design prevention and rehabilitation programs, or even performance training.

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