Wearable Technology (2022) Volume 3 Issue 1, *pp.* 39–46.
doi: 10.54517/wt.v3i1.1658

Asia Pacific Academy of Science Pte. Ltd.

ORIGINAL RESEARCH ARTICLE

Kinematic and kinetic analysis of transfemoral prosthesis

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ABSTRACT

The feasibility of using transfemoral prosthesis Otto bock with 3R80 knee and articulated ankle1C30 “Trias” was analyzed from the perspective of dynamics and clinic. The kinematic and kinetic study of gait were performed on 5 amputated volunteers and 5 controls using videography techniques and force platform. Kinetic asymmetry gait is one of the main causes of hip joint degeneration. Combining kinematic and kinetic variables, we can draw important conclusions related to the dynamic imbalance of the main causes of hip degenerative diseases through the clinical trials of radiography film and density measurement, which has become an important tool to evaluate the feasibility of prosthetic design.

Keywords: transfemoral prosthesis; kinetics; osteoporosis; osteoarthritis

1. Introduction

In patients with transfemoral amputation, the use of prosthetics will significantly change the biomechanics of their musculoskeletal system, consisting of its tendency during daily activities to recharge their intact limb, all of which can determine the occurrence of related diseases, such as knee and hip osteoarthritis of healthy limbs[1–5]. In most cases, insufficient mechanical stimulation associated with the bone remodeling process of amputated long bones can lead to osteoporosis and subsequent osteoporosis[3,4,6,7].

Scientific research in this field is increasingly related to the altruistic purpose of designing these devices[2, 8–13]. The ideal solution of this method is to study these design skills and methods. No wonder some researchers[5,14–17] have linked traditional terms such as comfort, mobility, mechanical strength and durability to comprehensive criteria to evaluate the suitability of prosthetic design. The authors believe that the functionality of lower limb prosthesis is usually related to the functional and well-being needs of amputees. This kind of well-being is mainly related to the gait pattern as close to the healthy limb as possible. with a minimal energy consumption and with the absence of disease in the residual joints caused by the prosthesis during the gait regimen.

In this study, we propose a method to evaluate

ARTICLE INFO
Received: December 29, 2021 | Accepted: January 31, 2022 | Available online: February 15, 2022

CITATION
Broche-Vázquez L, Sagaró-Zamora R, Ochoa-Díaz C, et al. Kinematic and kinetic analysis of transfemoral prosthesis. Wearable Technology 2022; 3(1): 39–46.

OTTO BOCK transfemoral prosthesis, which combines kinematic and dynamic gait studies of amputated volunteers and control group; combined with clinical Radiography and density measurement, important conclusions related to prosthesis function were obtained. For this purpose, the temporal and spatial variables of gait, kinematic and kinetic patterns (moments, forces and joint energy consumption), limbs (amputated and non-amputated) and the same number of pattern subjects of 5 amputated patients were measured and deduced. Finally, these results are clinically treated by Radiography technology to find signs of limb osteoarthritis that may be related to amputation and ambulation of amputees[1,18], and densitometry, a technique that allows mineral density to be measured (Bone mineral density, BMD)[19].

2. Materials and methods

2.1. Subjects

In the two groups of subjects, the amputated volunteers and the control group were characterized by gait. The transfemoral amputated patients used Otto Bock prostheses, with keen modal 3R80 and flexible foot modal 1C30 “Trias”, were 5 men with an average age of 32 ± 2 years, height of 1.75 ± 0.09 meters, weight 80 ± 10.97 kg. The control group consisted of five men whose age, weight, height and eating habits were similar to those of amputees. None of the subjects (patterns and amputees) had orthopaedic, neurological, cardiovascular or respiratory problems prior to the study. Before the test, all prostheses were thoroughly inspected, and the adjustment of each joint and the correct alignment of the prosthesis were checked. None of the subjects had any discomfort with their health and residual limbs, such as joint pain, motor stiffness, ligament instability, etc. The characteristics of volunteers are shown in **Table 1**.

2.2. Protocol

During the study, the subjects wore dark tight clothes and fixed reflective markers (0.02 m in diameter) affixed to the left and right iliac crest, greater trochanter, femoral epicondyle and lateral malleolus and L5-S1. For patients on the amputated side, according to Helen Hayes’s labeling scheme[21,22] (see **Figure 1**), the reflection point of the preset area (center of gravity, joint or other signal to obtain the required characteristics) is estimated according to the corresponding position on the healthy limb. Each volunteer developed a walk of about 6 meters for the measurement, with a free cadence of 20 repetitions.

Table 1. Characteristics of the amputation volunteers included in the study

Volunteer	Age (years)	Amputation year	Daily use (hours)	Dimension (m)	Weight (kg)
1 (I)	28	8	10	1.72	80
2 (I)	32	14	8	1.79	82.5
3 (D)	30	14	10	1.83	72
4 (D)	47	16	6	1.73	81
5 (D)	65	15	8	1.78	87
Average value	58	13	7	1.77	80

I: Left leg amputated
D: Right leg amputated

Figure 1. Location of reflective markers for gait analysis based on Helen Hayes marking protocol[21,22].

2.3. Material

These studies were conducted at the Gait Analysis Laboratory at the University of Brasilia (UNB) (see **Figure 2a**). The gait cycle is recorded through the professional software package Qualisys Motion Capture System in Gothenburg, Sweden, and then the data is transmitted to the computer. The processing of mark identification gear is carried out through the Qualisys Track Manager package, QTM, Gothenburg, Sweden[23]. It is an analysis software tool for managing and reporting video data. Together with high-speed motion video, QTM provides an advanced and accurate solution for biomechanical motion analysis (**Figure 2b**).

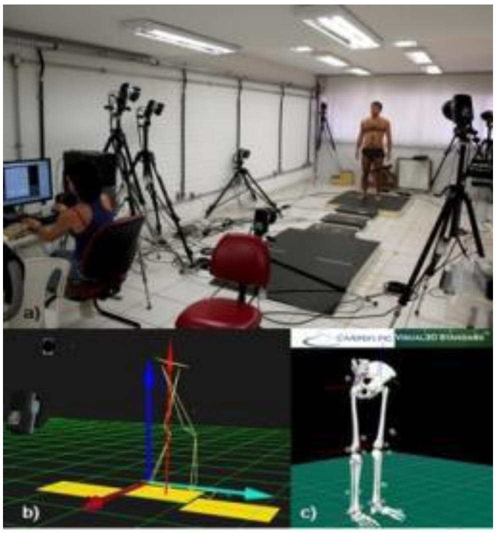


Figure 2. (a) Gait Analysis Laboratory at the University of Brasilia. **(b)** QTM software environment. **(c)** Conventional model of the lower limbs in visual 3D software (C-Motion).

After preprocessing, the data is exported to visual 3D v4 (C-Motion Inc., Germantown, MD, USA) for analysis and remaining gait processing. In visual3D, the traditional lower limb model is reconstructed (see **Figure 2c**), from which the reverse dynamics analysis is carried out to determine the angular displacements, torques and forces of the joints.

The data collections of amputees and healthy people were based on sagittal and frontal planes, which show the maximum displacement[24–26], as well as the values the components of the reaction force of the floor (vertical and anteroposterior) to facilitate the comparison between both limbs and the asymmetry of gait cycle.

2.4. Clinical trials of densitometry and radiograph

The bone densitometry results report provides the average bone mass values for each scan and measurement area, and uses digital and color images to link these average values with normal values based on the patient's age and gender. It is the main diagnostic tool for osteoporosis, so the risk of fracture can be determined.

Test results are usually reported as "T-scores".

The prevalence of osteoporosis and osteoporosis is estimated according to WHO classification. The test results are usually reported as "T-score" using the database referring to young people[27] (normal: T-score ≥1, bone mass reduction −1 to −2.5, and T score lower than −2.5 indicates osteoporosis.

According to clinical practice, the routine radiography examination was carried out for these cases, and all patients were examined. The technique used was to compare the anterior hip with the lateral femoral neck[28].

3. Results

3.1. Gait analysis

Table 2 shows the spatiotemporal variables generated by gait analysis of non-amputees and amputated subjects.

Table 2. Spatiotemporal variables of non-amputated and amputated subjects

Variable	Non-amputated subjects		Amputated objects			
			Complete state		Otto bock prosthesis	
	Average value	From	Average value	From	Average value	From
Operating speed (M/s)	1..177	0.319	0.986	0.281	0.986	0.272
Cycle length (m)	1.327	0.027	1.049	0.318	1.013	0.319
Step length (m)	0.670	0.023	0.579	0.021	0.653	0.039
Support time (s)	0.872	0.024	0.829	0.024	0.752	0.021
Rolling time (s)	0.455	0.022	0.418	0.013	0.501	0.012

From the spatiotemporal variables of the subject patterns of the two prosthesis types, it can be seen that there are significant differences in the spatiotemporal variables of the subject patterns of the two prosthesis types compared with the intact and amputated limbs of patients with transfemoral prosthesis. From the results obtained, it can be concluded that the speed and length of gait cycle of amputee subjects are lower than that of standard subjects. In this sense, the length of the steps is somewhat similar to the pattern.

Figure 3 summarizes the angular displacement of ankle, knee and hip joints, as well as the behavior of joint moments and energy consumption of two groups of subjects (patterns and amputees).

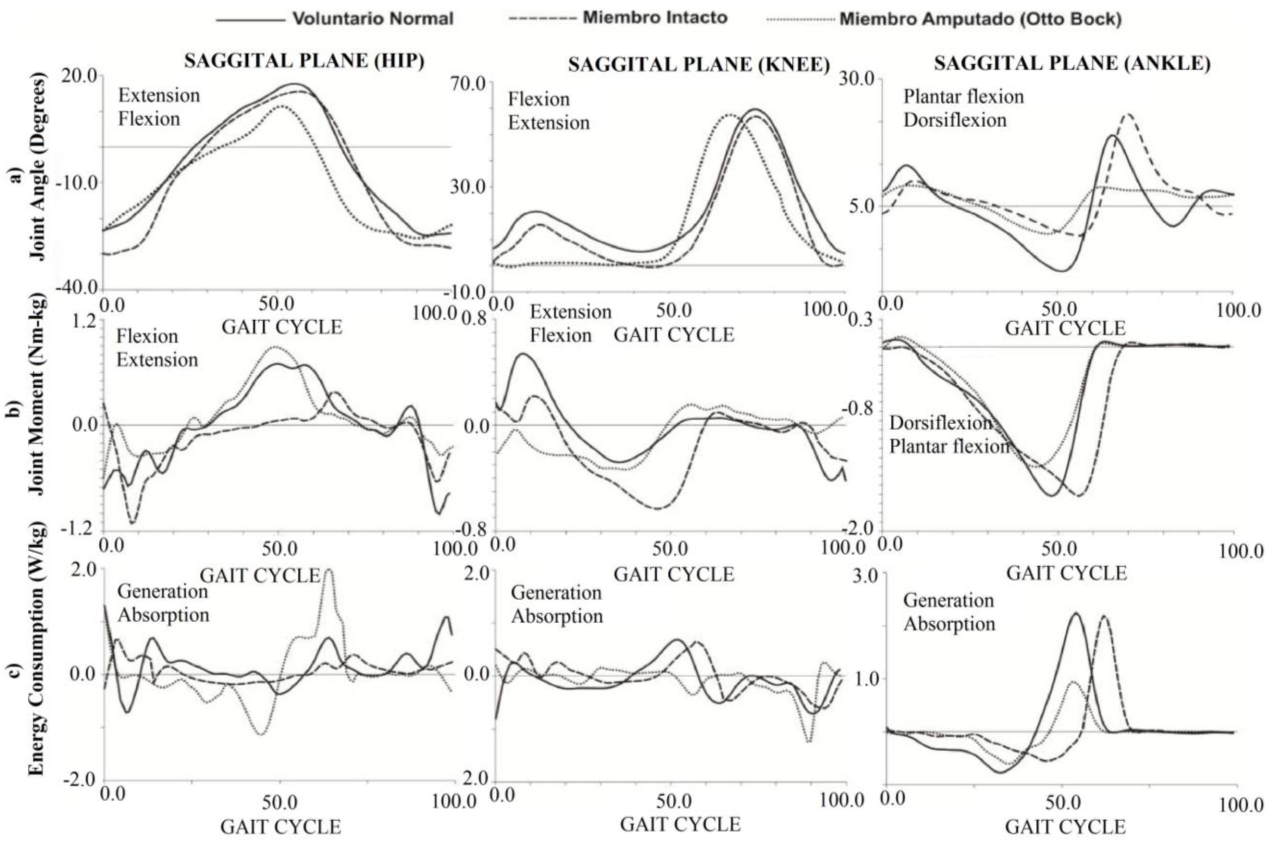


Figure 3. Angular and kinetic models for the standard subjects, intact and amputated limb of the transfemoral amputees. **(a)** Angular displacement of hip, knee and ankle. **(b)** Joint moments. **(c)** Energy expenditure.

The angular displacement of ankle joint has a pattern composed of an initial peak for normal people. In the standard subjects, the plantar flexion reaches about 10°, passing to dorsiflexion of about 15° at 53% of the cycle; finally, during the swing phase, the new plantar flexion of about 16° (68% of the gait cycle). In the intact limb of the amputee, the plantar bending is slightly smaller (5°), and then passes to dorsiflexion with a peak of about 7–8°, which has a certain delay compared with the control group. In the swing stage, there is a certain delay compared with the control group, experiencing a new plantar flexion, with a peak of 20° over 72% of the gait cycle.

In the amputated limb, the general spatiotemporal pattern includes a behavior of plantar flexion (≈ 5°) that is very similar to the that of the intact limb, and the posterior dorsiflexion also of the same amplitude as the intact limb, but that occurs only at 50% of the cycle, immediately transitioning to plantar flexion with a constant amplitude of only about 5° throughout the swing phase.

The angular displacement of the keen both for the standard subjects and for the intact limb of the amputees showed two important peaks. In pattern subjects, the first peak appeared in the support phase, accounting for 15% of the gait cycle, with an amplitude close to 20°, and the second peak appeared in the swing phase, accounting for 57°at 72% of the cycle. The intact limbs of amputees have similar behavior to some extent, with a certain delay and a slightly smaller amplitude.

However, the amputee's knee remained extended for almost the whole support phase, and then flexes at the same amplitude as the intact limb in the swing phase, but at 65% of the cycle, confirming that the amputee relied on the healthy limb for the longest time during walking. Without this slight knee flexion, dynamic shock increases.

As for the standard subject's hip joint, it showed initial flexion during the support of about 20°, and then extended for a long time for the rest of the support phase, with a peak extension of about 15°, and then was in a flexion state during the swing at the end of the cycle (80%). The amputee's intact limbs have similar behavior.

However, there are some differences in the behavior of the patterns and the healthy limbs of amputated hips. Although the initial flexion is no different from these, it maintains a longer support period during flexion and then extends to 8°. The time lag (top 10% of Health) confirms the argument that these patients rely on healthy limbs most of the time. The angular range of the displacement of the amputated hip joint relative to the healthy hip joint is significantly reduced, which is related to the temporary compensation mode due to the amputated valgus in the swing stage, so as to avoid rubbing the floor with the foot in the middle of the stage. The low angle motion of the hip joint is related to this and to the greater lateral swing of the trunk.

With regard to joint moment, for the ankle joints of standard subjects, they showed a small dorsiflexion moment at the beginning of support, which lasted for a short time, about 0.1 N.m/kg, and quickly transformed into a plantar flexion moment with a peak value of 1.6 N.m/kg (50% cycle), which gradually decreased to zero at 60% of the cycle.

The joint moment of healthy limbs of amputees showed a very similar pattern, except that the peak plantar moment with the same amplitude as that of standard subjects appeared at 60% of the cycle, that is, there was a certain delay compared with standard subjects. In contrast, the amplitude of the prosthetic limb was low (1.4 N.m/kg), with a delay of about 18% compared with the healthy limb.

The behavior pattern of knee moments showed that the extensor moment of the healthy subjects was 0.5 N.m/kg, which was slightly higher than that of amputated patients with intact limbs (0.2 N.m/kg), which showed a certain delay. The next peak, this time, was more pronounced for the amputated intact limb than for the control volunteers (0.65 N.m/kg and 0.2 N.m/kg, respectively). In both cases, the subsequent extended behavior has very similar behavior, gradually approaching zero.

However, the amputated limb shows a behavior different from the standard and intact limb, which is easy to understand from the angular displacement of the prosthetic knee joint. As observed, the flexor moment generated by knee hyperextension is logical, and then the flexor moment is increased to 0.3 N.m/kg, which is higher than that recorded by standard subjects, but much lower than that generated by the intact limb (0.653 N.m/kg), which is an important reason for the gait asymmetry of amputees.

With regard to the pattern of hip moment, the extensor moment was 0.3 N.m/kg in 16% of the cycle, then became the flexor moment with a peak of 0.7 N.m/kg (50% of the cycle), then gradually decreased, and the pattern was repeated at the next heel strike. At the beginning of the support, the intact limb has an extension peak of 1.2 N.m/kg, which gradually becomes a flexion moment at the end of the support phase (maintaining the extension for most of the support phase) with an amplitude peak of only 0.3 N.m/kg.

During heel support, the hip joint of the amputee is stable at about 0.3 N.m/kg, and is transformed into a flexor moment with a peak of 0.9 N.m/kg at the end of the support stage to gradually reduce.

The uneven kinematic and kinetic behaviors between the amputated and non-amputated limbs reflect the compensation process in the process of amputated walking, showing obvious asymmetry.

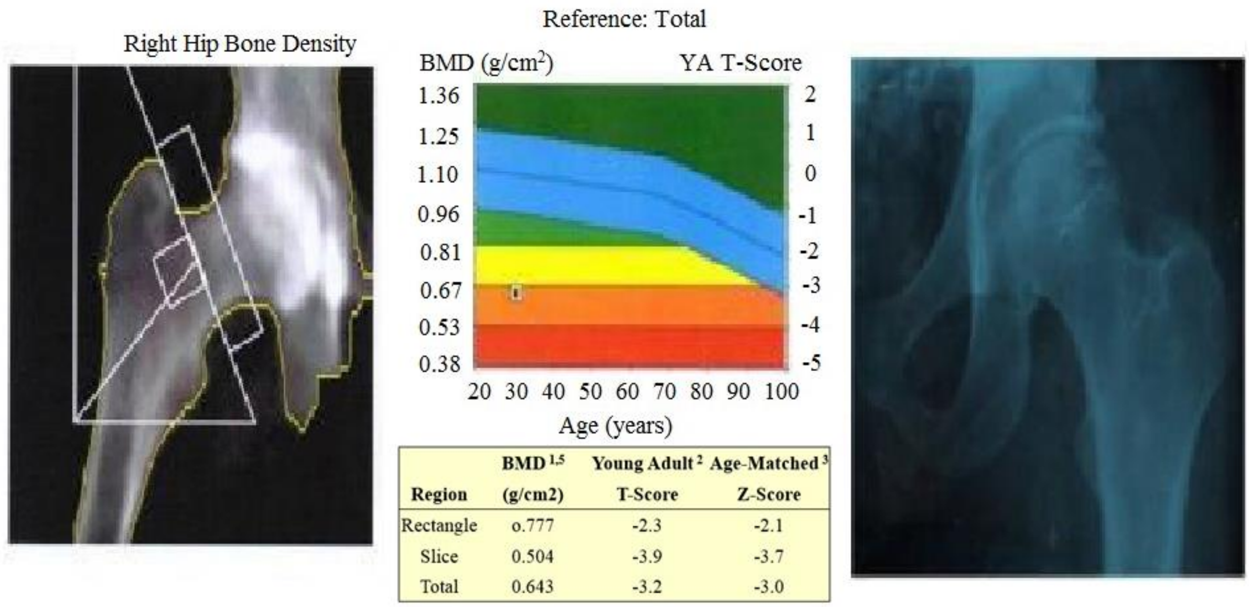
In terms of energy consumption, the results show that the mode of ankle joint is different in peak amplitude and development time, which is consistent with the above kinematic and dynamic behavior explained. The peak energy absorption of intact limbs of standard subjects and amputees corresponds to the value of 0.7 W/kg. The only difference is that the latter occurs in 50% of the cycle, that is, there is a certain delay compared with standard subjects. This behavior has similar characteristics at the end of support, because they produce a peak of 2.2 W/kg, and the delay of intact limbs of amputees is 10° (≈ 60% of the cycle). At the end of the support phase, the amputated ankle produces a peak of only 1W/kg, corresponding to the smallest plantar flexion.

For the knee joint, the resulting energy consumption showed a very similar pattern for the standard subject and the intact limb of the amputee with peak value very similar to heel shock and delayed -off for the intact limb in about 0.3 W/kg and 0.8 W/kg. For the amputated limb, when supporting, taking off and swinging, the power generation is low and the consumption is large.

As for the energy consumption of the hip joint, it is characterized in that the subjects control the two power generation peaks corresponding to heel support and take-off. For the intact limb of the amputee, during the heel support process, the peak is consistent with the pre-formed pattern of the subject within a certain period, and is delayed with the take-off of the foot. For the hip joint of amputated patients, the absorption and energy production levels are always much higher than those of standard subjects and intact limbs of amputated patients, with peaks of 1 and 2 W/kg.

3.2. Clinical study of bone mineral density measurement and radiography examination

The study was conducted in amputated volunteers (see **Figure 4**). Only 2 of them developed osteoporosis during amputation, and in no case did they find clear joint damaging osteoarthritis in healthy limbs.



Region	BMD 1,5 (g/cm2)	Young Adult 2 T-Score	Age-Matched 3 Z-Score
Rectangle	o.777	-2.3	-2.1
Slice	0.504	-3.9	-3.7
Total	0.643	-3.2	-3.0

Figure 4. (**a**) The density measurement of a 30-year-old and 14-year-old amputee showed that the amputee had obvious osteoporosis. (**b**) The radiography of the patient's non-amputated limb showed a significant lack of joint injury.

4. Discussion

The recording results of support and balance time show that during walking, amputee volunteers support their intact limbs longer than amputee volunteers, which has been reflected by other research work[9,12,13,29], and which is a clear sign of gait asymmetry and affects the pathological detection of amputee volunteers' intact limbs and stumps[4,6,30]

In general, all joints of the intact limbs of amputated volunteers describe trajectories similar to those of control volunteers, although at some peaks, the delay time is longer and the amplitude is lower, which has been reported by the authors[8,31,32] and other researchers[33–35].

The prosthetic knee remains extended throughout the support phase to bend during swing, but with10 degrees advance to the healthy one, which confirms the shortening of support time. Without this slight knee flexion, the dynamic shock increase.

As shown in the figure, the results of densitometry and radiography examination show that the asymmetry of transfemoral prosthesis is largely the cause of premature lesions in both limbs. However, in some amputated volunteers, the occurrence of osteoporosis and osteoarthritis depends on dynamic factors affecting the process of bone reconstruction, such as gait speed, step length, especially during support and swing time.

Although the literature[4,35] mentioned the presence of osteoarthritis in non-amputated limbs, no conclusion has been drawn in this study because the amputated patients studied did not have this pathology. A study involving more patients will help determine the extent to which the prosthesis is involved in these diseases.

5. Conclusions

The gait analysis of amputees showed that there were significant differences in gait patterns between healthy limbs and amputees compared with healthy patterns, and only prosthetic volunteers related to support and swing time had significant differences, which confirmed that amputees leans longer on healthy limbs than on the amputee during walking.

Gait research provides a very important detail for amputees using 3R80 knee joint, that is, the significant abduction of the amputated limb and the corresponding pelvic inclination.

The results of densitometry and radiography examination can determine the changes of mechanical stimulation in different areas of the two limbs and estimate the presence of osteoporotic areas in the absence of mechanical stimulation. However, in these dynamic stimuli, the effects of variables such as support and swing time and walking speed need to be considered. Combined with the methods used here, we can establish a standard of prosthetic function, which is considered to be a method of artificially designing close to healthy limb function without involving the occurrence of corresponding diseases.

Conflict of interest

The authors declare no conflict of interest.

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