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DEVELOPMENT OF A PASSIVE PROSTHETIC ANKLE WITH SLOPE ADAPTING CAPABILITIES

Sandesh G. Bhat
Arizona State University
Mesa, AZ, U.S.A.

Sangram Redkar
Arizona State University
Mesa, AZ, U.S.A.

Thomas G. Sugar
Arizona State University
Mesa, AZ, U.S.A.

ABSTRACT

In this work, the authors first study different designs of prosthetic ankles. Then, a new design is proposed, and its dynamics are discussed. Various experiments are conducted to verify the concept. The results of the experiments are discussed, and a conclusion is drawn based on the discussion. An OpenSim simulation is run to emulate the effects of the prosthesis on an amputee's residual leg. Further iterations of the same design are then discussed and presented. Finally, a conclusion is drawn on the usability of the ankle along with some suggestions for future research.

INTRODUCTION

The field of prostheses has advanced immeasurably over the past decade. The innovations in the field of motor control, control algorithms, and machine learning have all aided in the development of newer and more efficient prosthetic devices. But all these innovations have been limited to powered prosthetic devices. These active prostheses are usually expensive and out of reach for a very large population with low income. Amputations occur mainly because of battlefield accidents. Also, the United States is seeing a rise in diabetes and diabetes related amputations. In 2016, a reported 29 million Americans were suffering through diabetes and 86 million were diagnosed with prediabetes [1]. The need for a cheaper alternative to powered prostheses is evident.

Passive protheses are cheaper compared to active prostheses. Hence, they are a better alternative for low income groups. The field of passive prosthetic devices has seen very few advances. The newer passive prosthetic ankles are still inefficient, and the user must undergo a lengthy and arduous learning period to get accustomed to their prostheses. This demands a better prosthetic ankle with a simple design and cheap manufacturability that also provides a better experience for the user. The authors developed such a prosthetic ankle in [2]. In the following article, the authors revisit this prosthesis and discusses experiments done on the same while also presenting the next iteration and discussing experiments conducted to verify this new concept. The authors have also submitted a journal paper on the same topic which is pending review.

DESIGN

The CAD render in FIGURE 1 is from a master's thesis which can be found in detail in [2]. The project is also under review as a journal paper (ASME Journal of Medical Devices manuscript no. MED-18-1034) The design consists of two springs and two holders with a carbon fiber foot attached to act as a leaf spring. The spring in the front is at an angle corresponding to the Anterior Tibial tendon angle of about 37 degrees. The tendon angles are taken from Procter et al. [3]. The rear spring is vertical and acts as the Achilles tendon providing support while the user stands still. As can be observed in the CAD model in FIGURE 1, the design is simple and only uses five pins to hold everything together. The pyramid on top is used to attach the prostheses to the amputee's residual leg. The prosthesis was designed and optimized for a subject weighing 80 kg.

All the designed parts were analyzed using Finite Element Methods for stress characteristics. The parts were loaded with 2.5 times the intended load to ensure safety. All the parts were safe in the analysis. The carbon-fiber foot was selected based on provided data, so it wasn't analyzed.

Two prototypes of the above design were manufactured for testing in different conditions. One of them was prototyped for a healthy human being weighing 80kg aka. Full weight tests, while the other prototype was used for tests conducted on a robot aka. Reduced weight tests. Both these tests are discussed in the next section.

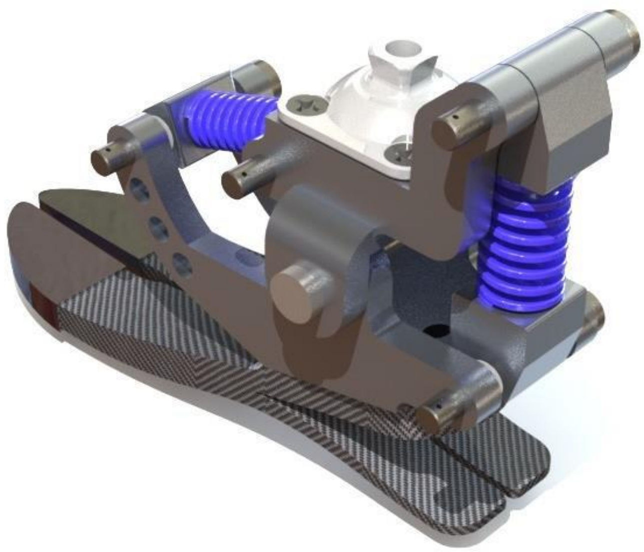

FIGURE 1. CAD OF PROPOSED PASSIVE PROSTHETIC ANKLE [2]

EXPERIMENTS AND RESULTS

The first experiment was conducted on a Baxter robotic arm. The prosthesis was attached to the arm using a custom-built intermediate piece as shown in FIGURE 2. The Baxter robot was programed to emulate human gait on a force plate. The force plate was capable of reading forces only in the normal direction. The robot was programmed to apply its maximum payload (5 lbs./ 2,2 kg) as an emulated body weight on the prosthesis.

FIGURE 2. EXPERIMENTAL SETUP USING BAXTER ROBOT [2]

The experiment was conducted as 4 sets of 8 steps each. Ground Reaction Force (GRF) data from these sets were collected, normalized, and plotted with a reference data from Michael W. Whittle's book [4]. FIGURE 3 shows the plot for comparison of the said data. The plot of the experimental data from the tests were averaged over 16 steps. These 16 steps were chosen based on low noise criteria. The experimental data showed a similar peak value as the reference data while also displaying two distinct peaks. These results were analyzed, and it was concluded that the GRF data was satisfactory and displayed a potential in the current design. Hence full body weight tests were conducted.

The second experiment was performed on an able-bodied subject wearing a crutch to simulate amputee gait. The prosthesis was attached to a crutch, as shown is FIGURE 4. This setup was used by an able-bodied subject while walking on an instrumented treadmill with a harness attached to the subject to ensure safety. It was ensured that the harness did not decrease the subject's body weight on the instrumented treadmill that measured the subjects GRF while performing the experiment. A motion capture system tracked the subject's motion and stored information on the subject's gait. The GRF data in conjunction with the motion capture data were used to calculate data such as the ankle angle and ankle moment. The experiment was performed for multiple repetitions with 30 seconds of walking, each, at 1.2 m/s and out of these repetitions, 16 steps were chosen that had the minimum noise. These data were post-processed and plotted as can be seen in FIGURE 5 and FIGURE 6.

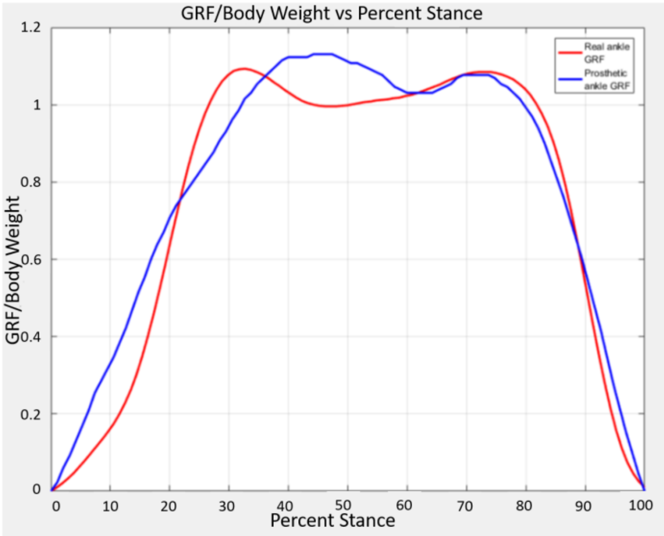

FIGURE 3. COMPARISON BETWEEN GRF OF ACTUAL HUMAN GAIT VS. SIMULATED GAIT [2]

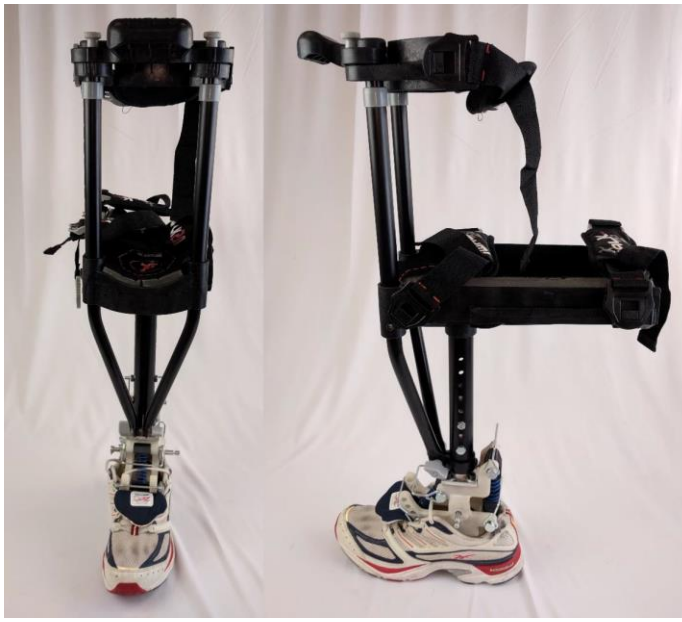

FIGURE 4. TESTING RIG FOR SIMULATED AMPUTEE GAIT TESTS [2]

The data shown in FIGURE 5 is a comparison between the ankle angle data from the experiment (solid black line) and the reference data (dashed black line) taken from M.W. Whittle's book [4]. The experimental data has a similar profile in the beginning but deviates from the reference data towards toe-off. The absence of an external force led to the prosthesis ankle angle stopping at zero. With a negative ankle angle around toe-off, a higher push-off force could have been achieved. But due to the passive nature of the prosthesis, this was not viable.

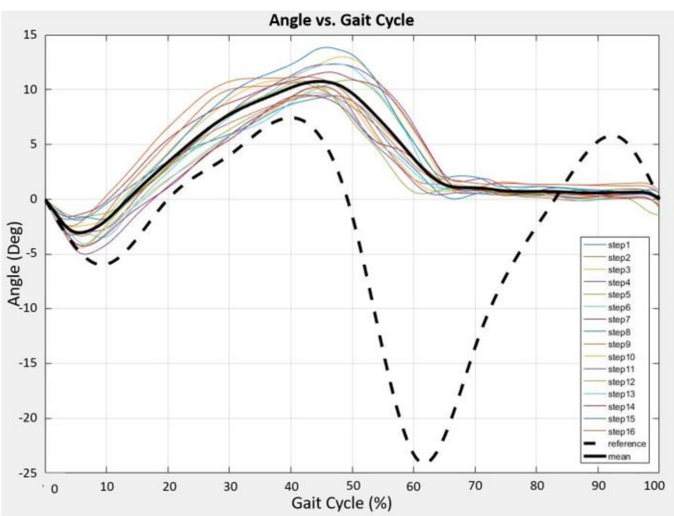

FIGURE 5. ANKLE ANGLE VS GAIT CYCLE FOR THE ABLE-BODIED SUBJECT TEST (SOLID LINE IS THE MEAN OF ALL THE ANGLE DATA COLLECTED, DASHED LINE IS THE REFERENCE ANGLE CURVE TAKEN FROM M. WHITTLE) [2]

Along with the prosthesis ankle angle, moment data was also collected and analyzed from the experiment. FIGURE 6 shows a comparison between the experimental data with the moment data from M.W. Whittle's book [4]. The data from the experiment consisted of a lot of noise which was reduced using a smoothing algorithm. The moment of the prosthesis rises right after heel strike. This rise can be attributed to the experimental setup. The crutch used by the subject allowed for a low range of rotation for the knee. Due to this hindrance, a small limp was noticed in the subject's gait. This caused the moment at the ankle to rise rapidly. The negative moment seen after toe-off was an anomaly. The most probable reason for this anomaly was noise due to the oscillation of the springs. The smoothing algorithm corrected the noisy data, but the final output was a negative moment. To fix this issue, a possible solution would be to use dampers along with the springs, but the required spring stiffness would have to be recalculated and reoptimized.

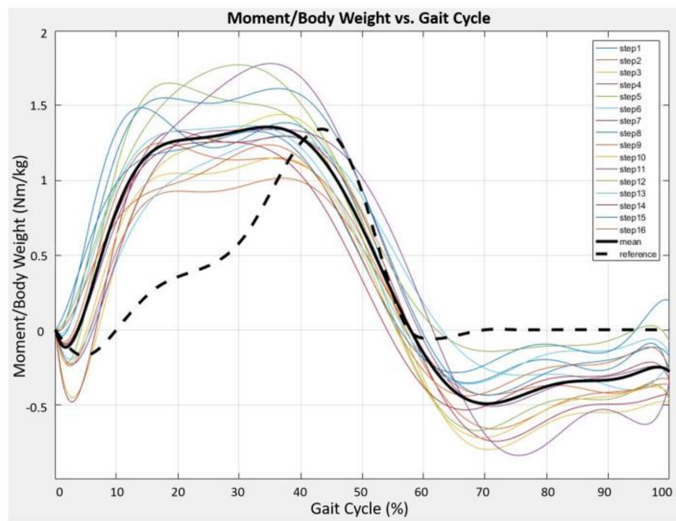

FIGURE 6. MOMENT PER BODY WEIGHT VS GAIT CYCLE FOR THE ABLE-BODIED SUBJECT TEST (SOLID LINE IS THE MEAN OF ALL THE ANGLE DATA COLLECTED, DASHED LINE IS THE REFERENCE ANGLE CURVE TAKEN FROM M. WHITTLE) [2]

All the data collected from the experiments represented the output of the ankle purely in a mechanical sense. The bio-mechanical aspect of the prosthesis still needed verification. It was wished to see how the prosthesis would affect the user and their body in long term. The next section discusses the attempts made to understand the user's interaction with the prosthesis.

BIOMECHANICAL SIMULATION

To quantify the effects of the prosthesis on an amputee's residual leg, an OpenSim simulation was performed. For this simulation, a model of an able-bodied subject from OpenSim was utilized and modified to incorporate the prosthesis. The model was changed to represent a trans-tibial amputee. Annex A consists of the link to the code along with the modifications. All the credits and due acknowledgements are contained in the code. Further details regarding the unmodified model can be found in Delp et al.'s article [5].

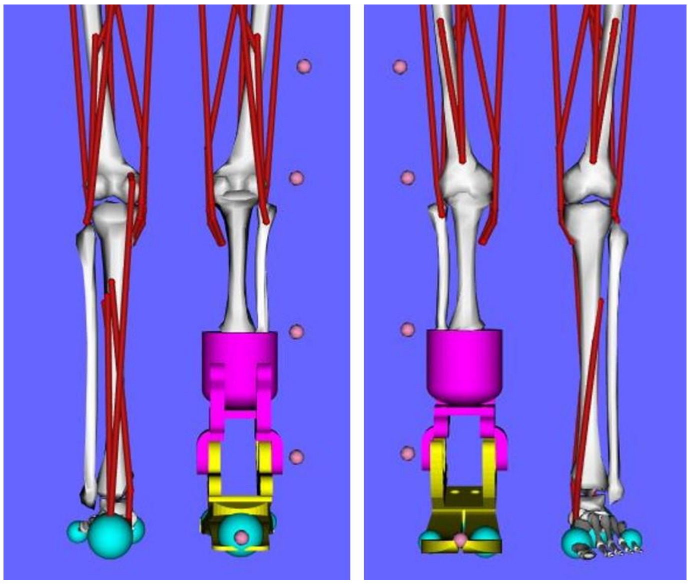

FIGURE 7. BACK AND FRONT VIEW OF THE MODIFIED OPENSIM MODEL WITH THE MARKER LOCATIONS MARKED (ORANGE DOTS) [2]

FIGURE 7 shows a part of the model used in the simulations. The model also consisted of the motion capture marker data. This was used to connect the motion performed by the subject during the experiments to the model directly. The simulation was run, and data regarding the force delivered by the tendons were acquired. These data were plotted as seen in FIGURE 8. The four lines on the plot represent the Biceps Femoris Long Head (BFLH) and the Vastus Intermedius (VI) for the right and the left leg. By analyzing the plot, it can be deduced that the right BFLH (blue) is a little under actuated when compared to the left BFLH (red). Also, the right VI (pink) does not go through as much actuation as the left VI (green). It could also be characterized by a limp that can be observed in amputees using ankle prostheses. This was partially because the crutch hindered the subjects knee motion. But it was also indicative of the muscles in the amputee's residual leg that would be used less. This might lead to muscle atrophies in the long term.

These test data were instrumental in deciding that the prosthesis was operational and acquiring a proof of concept. It was also interesting to note that the prosthesis only had one degree of freedom at the ankle and was operational only on level ground. For more than a single degree of freedom, the prosthesis would need to have a spherical joint for the ankle joint. The fixed hinge in the current design would not allow for any change due to the angle of the ground. Also, a spherical joint in place of the fixed hinge won't be good design choice for a passive prosthesis. A middle ground between the spherical joint and the fixed hinge was reached by the authors and this improved design will be discussed in the next section.

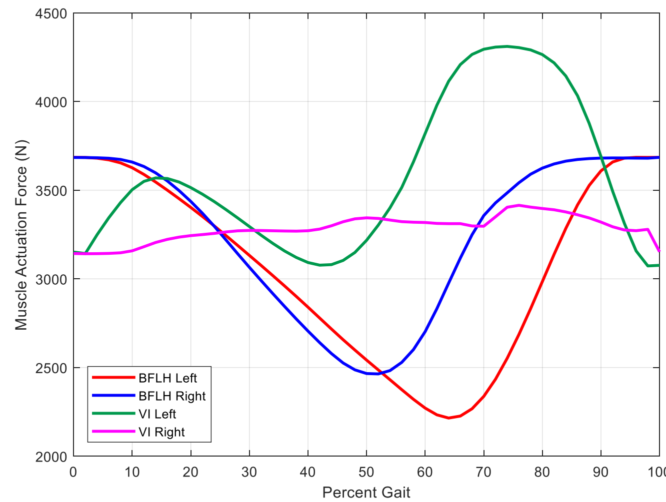

FIGURE 8. MUSCLE ACTUATION FORCE VS PERCENT GAIT

FUTURE RESEARCH AND DISCUSSION

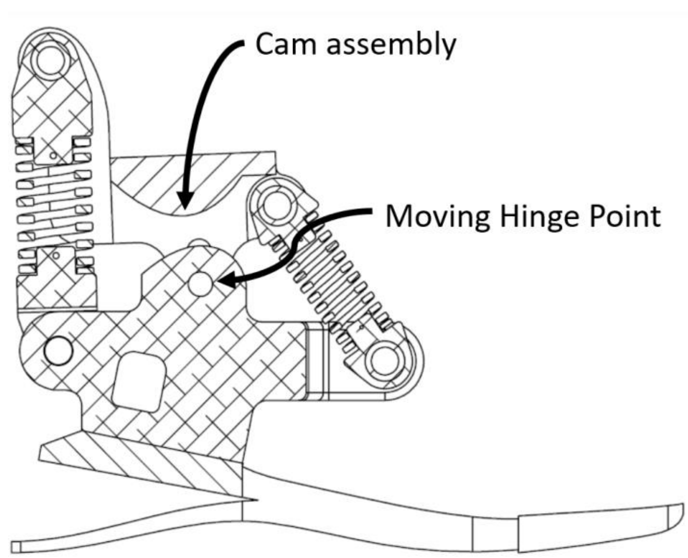

FIGURE 9. SIDE CROSS-SECTIONAL VIEW OF THE IMPROVED PROSTHETIC ANKLE. [2]

After the concept of the angled spring was confirmed to work, the design was further iterated to overcome its major limitation: slope gait. This was accomplished by modifying the hinge design to incorporate a cam mechanism. The cam surfaces were coated with a high friction material to make sure no slipping occurred during operation. The mechanism allowed the ankle to have a variable contact point for the ankle to "roll". The springs were still at the same angle as before. FIGURE 9 shows the cam mechanism in its cross-sectional view.

Three types of gait were considered: (1) Level ground gait, (2) Uphill gait, and (3) Downhill gait. During different types of gait, the users weight with the angle of initial contact with the ground made the two cams come in contact at different points with different configurations. This gave the mechanism its versatility.

The design was iterated further using trial and error method, and the cam profile was optimized. These iterations went through the robot testing mentioned before, and uphill (positive 12 degrees grade) and downhill (negative 12 degrees grade) gait simulations were added to the test. The tests were performed for 3 repetitions, 16 steps each. FIGURE 10 shows the average of the data collected for each type of gait. The data was plotted against percent stance as the GRF during the swing phase was zero. The plot of level ground gait showed a peak of 1.65 units normalized after heel strike, but the second peak lacked the magnitude. This was because of slipping in the hinge cam. The high friction material was unable to coat the surfaces properly. On the other hand, Downhill gait was a little better in that respect. The two peaks were pronounced, but the first peak lacked magnitude. It was believed to be because of the simulated gait, the robot lacked the response a human gait would have had on a downward slope. Hence, the plot was considered invalid. The plot of GRF for uphill gait did not prove to be satisfactory for the same reason. Hence, the modified prosthesis will be further tested for downhill and uphill gaits in the future.

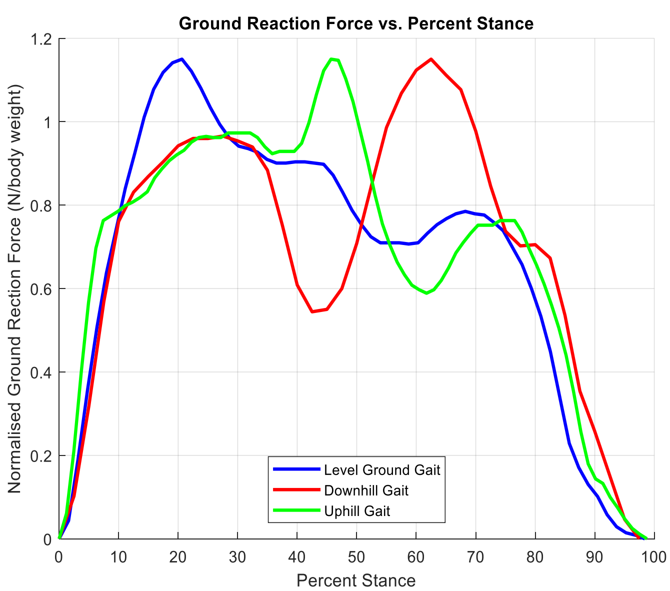

FIGURE 10. GRF VS PERCENT STANCE DATA FOR LEVEL GROUND, DOWNHILL, AND UPHILL GAIT.

The previous design was improved and a prosthesis capable of more than the original design was presented. The new design was tested, and a few shortcomings were found. The future iterations will incorporate changes to fix these shortcomings and a better testing strategy will be devised to ensure the results are reliable.

CONCLUSION

After analyzing all the experimental data, the authors believe the discussed design needs more tuning. A well-tuned prosthesis will result in a better user experience, lower user fatigue, and near perfect gait. The prosthesis was designed to fit the test subject and hence a proper understanding of the user's biomechanics to optimize the prosthesis would be required. Apart from this limitation, the authors believe the prosthesis has potential and would result in a more affordable and easy to maintain device for the user. The authors will continue to make improvements in the design. A world with near perfect prosthesis is within sight.

NOMENCLATURE

GRF: Ground Reaction Force.
BFLH: Biceps Femoris Long Head
VI: Vastus Intermedius

ACKNOWLEDGMENTS

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ANNEX A

OPENSIM SIMULATION MODEL

The code in the link below was originally taken from an example file included in OpenSim. The original credits for the code are included in the code at the beginning. This work can be distributed, tweaked, and built upon. The original authors must be credited in any such reproductions.

Link: https://gist.github.com/mrsandeshbhat/31a439f9ada3dda51b3d14ab8518ec37

FIGURE 11 shows the full body model used for simulation.

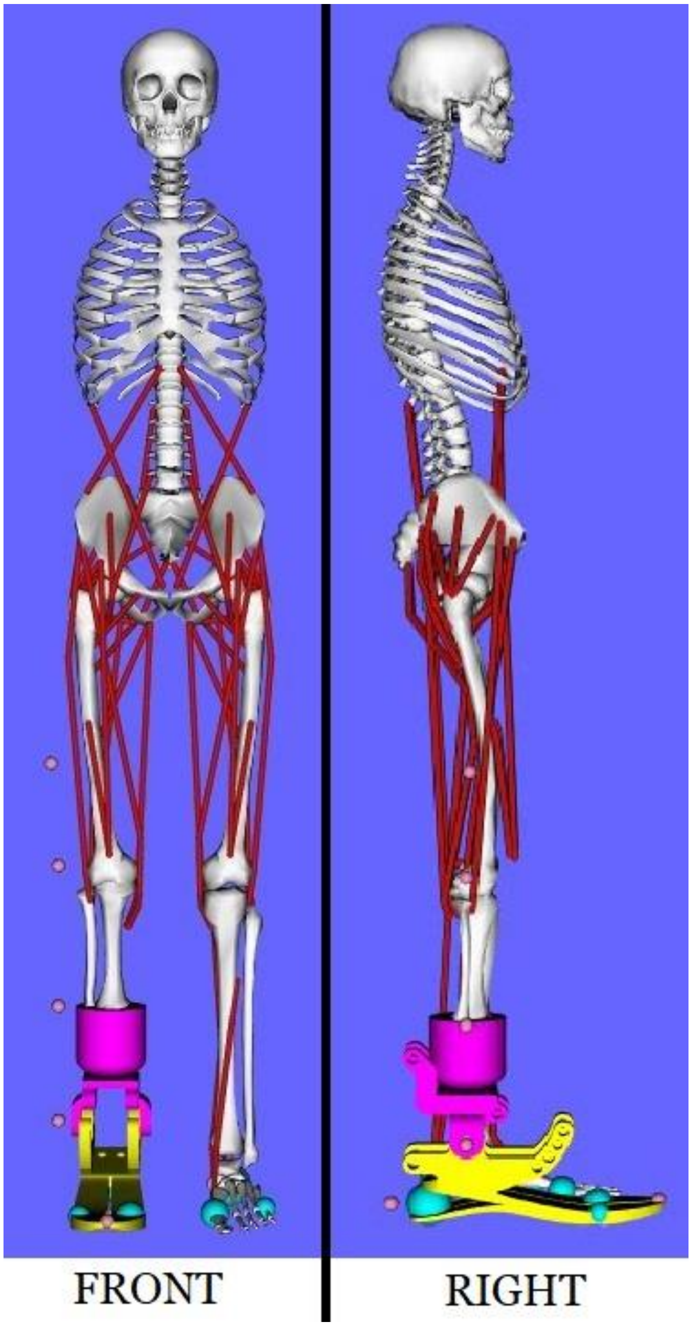

FIGURE 11. FULL SCALE MODEL USED FOR OPENSIM SIMULATION.