ASSISTIVE WALKER

SUBMITTED BY

AHMED SAMEER KHAN M ME11B123

ADDARSH CHANDRASEKHAR ME11B154

SHARA RHAGHA WARDHAN B ME11B121

OBJECTIVE

The project aims at developing an orthotic device which can serve as an assistive walker for people suffering from impaired control of the ankle. One of the predominant ankle gait abnormalities is the **Foot Drop** which is the inability of the tibialis anterior to perform concentric or eccentric action. In such cases, the orthotic should serve the purpose of functionally mimicking the important functions of the tibialis anterior in normal gait. The orthotic device attempts to replicate the normal gait kinematics as much as possible.

NORMAL GAIT

Fig.1 Normal Gait Cycle

In the normal gait cycle, the joint motions of the hip, the knee and the ankle are given by the following plots:

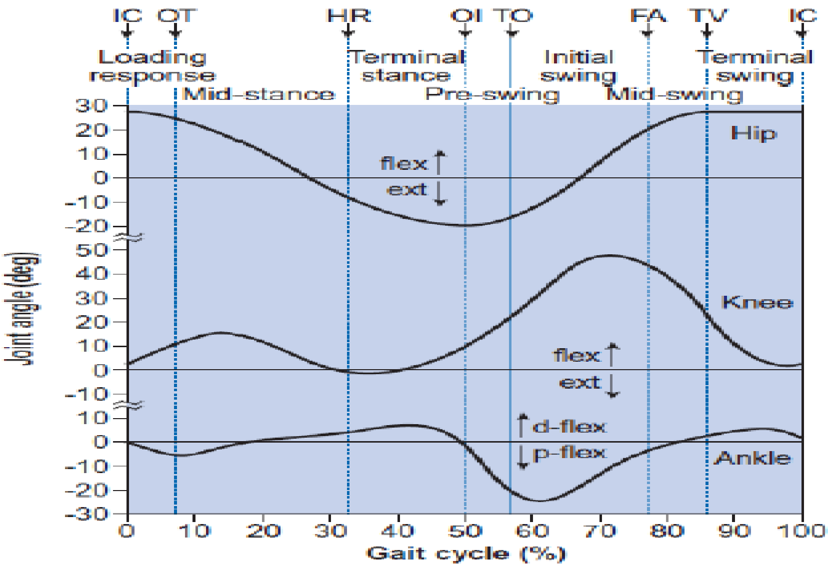


Fig.2 Joint Motions in knee, thigh and ankle

The role of the tibialis anterior is the primary concern. The two major actions are given by:

1. Eccentric action just after heel contact: Controlled shock absorption in the loading response phase of the gait cycle
2. Concentric action during swing phase: Dorsiflexion of the ankle during the swing phase so that the leg can clear the ground.

The introduction of the orthosis attempts to replicate the functionality of the tibialis anterior, mentioned above.

LOSS OF NORMAL GAIT

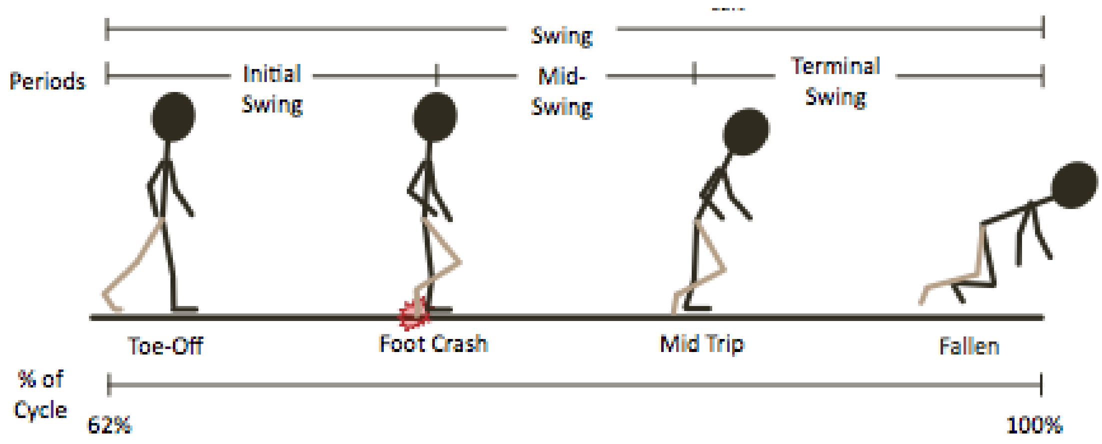


Fig.3 Consequence if normal gait is attempted with impaired control of tibialis anterior

In the case where the person has weak tibialis anterior, if the person attempts to walk in the normal gait cycle he will be unable to clear the ground as the foot would be plantarflexed in the swing phase. This happens due to the absence of the action of the tibialis anterior. More importantly after heel contact, foot slap will occur. Under normal gait, the tibialis anterior would have eccentrically damped this, but absence of its control causes the foot slap.

PATHOLOGICAL GAIT

There are ways in which the person can still walk by modifying his gait cycle at the expense of increased energy expenditure, lesser step length and increased duration of the gait.

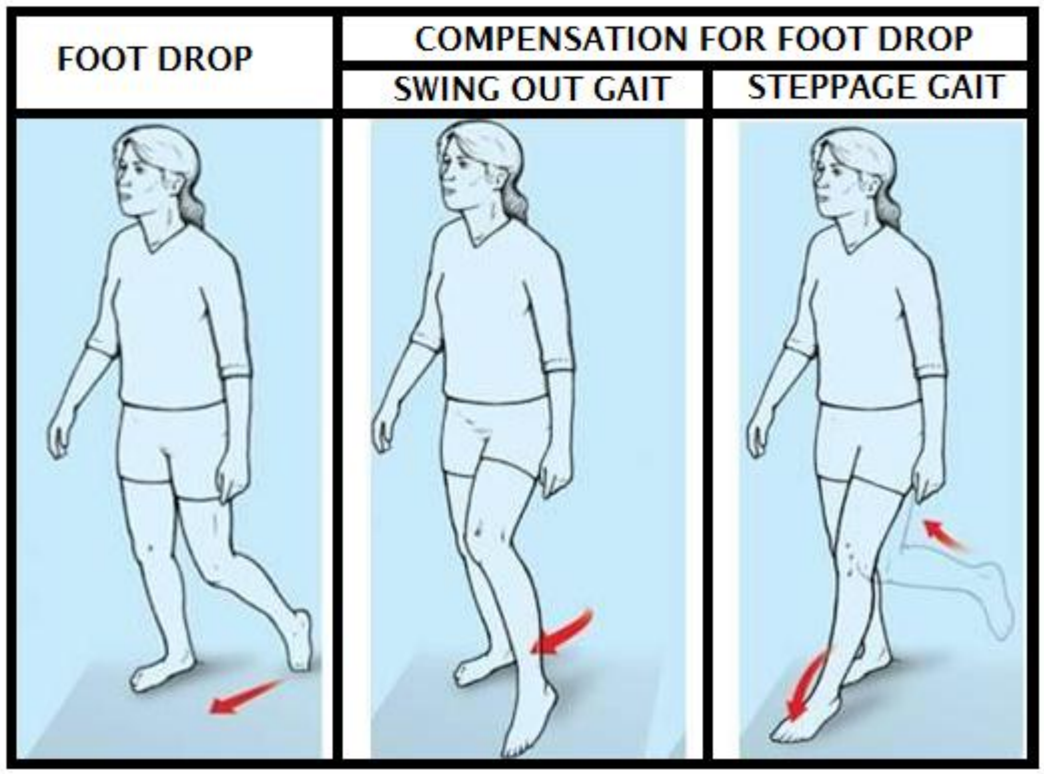


Fig.4 Abnormal Gait to adapt to the impaired control of tibialis anterior

STEPPAGE GAIT

During the swing phase, the person has to flex his hip and knee more to clear the ground. To reduce foot drop, the person tends to extend his knee lesser than normal thereby reducing the step length. So he cannot walk at the self-selected speed in which minimum energy is expended. Instead he will walk slower thereby increasing the duration of the gait cycle. Another adjustment is that instead of landing on his hell, he will land on his forefoot.

SWING OUT GAIT

In this abnormal gait, the person tries circumduction to clear the ground by flexing hip and abducting his leg. Here, the amount of energy expended is high compared to normal gait.

MECHANISM

The main requirement of the mechanism is to replicate the functional aspects of the tibialis anterior muscle which includes

a. Concentric Dorsiflexion during swing phase.

b. Eccentric Plantarflexion during Loading response to dampen the shock absorption.

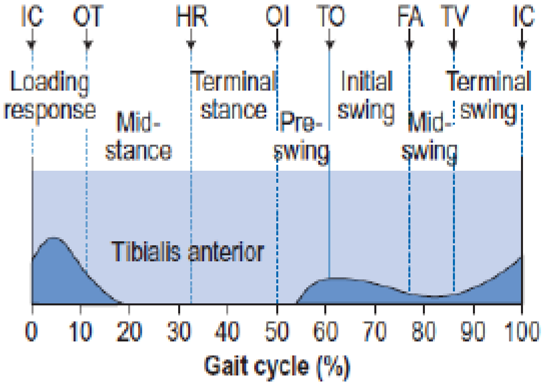


Fig.5 Major Muscle activity of tibialis anterior

The mechanism should be able to restore control of the ankle during swing and during heel contact so that these requirements are met.

Referring the joint motion diagram of the ankle(fig.1), the mechanism should be able to cause ankle joint motions as close to it as possible.

In order to know which foot orientation(angle) is to be achieved at each stage, some reference must be drawn from other joints. Based on the angle of flexion/extension at some other joint, the mechanism recognises which stage the cycle is in and the ankle is appropriately positioned. These can be thought of as INPUTS to the mechanism and the desired OUTPUT is the ankle's appropriate position.

The possibilities available for input are:

1. Angle between femur and tibia
2. Angle between hip and femur
3. EMG data of other muscles like the quadriceps, triceps surae and hamstrings.

Each of these inputs are captured mechanically/electrically through one of the following methods:

1. Gears
2. Cams
3. Linkages
4. Surface Electrodes
5. Flex sensors
6. Switches

POSSILE MECHANISMS

Some of the possible mechanisms combining the above are

1. Find the position of the centre of pressure using switches under the footwear and using this to actuate a piston whose end effect substitutes for the action of the tibialis anterior.
2. Use surface electrodes to get the data of muscle activity and use this data to manipulate a piston.
3. Use flex sensors at joints to convert the rotated angle into resistance and use this as an input to manipulate a piston substituting the tibialis anterior
4. Use linkages to know the orientation of the tibia and femur and use a mechanism to manipulate the position of the foot.
5. Use gears to convert the multi-direction rotation of joints to get a uni-directional gear and use this to construct a cam profile where the gait cycle’s ankle motion is distributed in the 360 degrees of the cam by some transfer function model.

Issues that could be faced with the above mechanisms are

1. Electrical components can fail and their maintenance will be high
2. Linkages will be very complex and may cause unnecessary weight
3. EMG data will be superimposed with huge noise data and filtering will be an issue.
4. Mechanical components may undergo wear and tear with time.

MECHANISM CHOSEN

A mechanical solution is sought because it is a lot easier to maintain compared to a sophisticated electrical system. Issues of powering the circuitry or connection issues are eliminated. The mechanism will derive some additional energy from the person to achieve the required end effect. But this effect should be dwarfed by the advantage of having overcome the problems of the pathological gait. If the net joint forces and reactions are closer to the normal gait cycle, then the compromise would be minimum.

A mechanism is initially sought after to replicate the kinematics of the normal gait. Once this is done, a force analysis is carried out, subject to certain simplifying assumptions, to estimate the new joint forces and reactions. The entire procedure has been coded in MATLAB so that data can be generated for all points on the GAIT cycle.

DATA OBTAINED FOR THE NORMAL GAIT CYCLE

1. Joint motions at the knee, hip and ankle
2. Angular velocities and angular accelerations
3. Joint forces and reactions in normal gait – to compare
4. Ground Reaction force (horizontal and vertical) and the location of centre of pressure.

PROPOSED MECHANISM

SOURCE OF INPUT MECHANISM:

A CAM-driven mechanism was considered due to its feature of **exact-function generation.** In simple words, if there is a certain follower output desired and the cam input angles are given corresponding to each output, the CAM profile can be designed.
In this mechanism, a follower and a CAM are provided. The relative angle between them changes through the gait cycle. So, in the translating reference frame at the centre of the CAM, the CAM would be rotating and the follower would be reciprocating like a regular CAM-follower mechanism.

The follower is connected by a series of linkages to the foot. So, given the desired foot motion(dorsi- and plantar-flexion) at various stages of the gait cycle, and given the corresponding CAM angle, an appropriate CAM can be designed.

INITIAL ATTEMPT:

The CAM is pivoted to a structure covering the leg and its point of pivot is made to almost coincide with the knee joint. There is a follower whose guide-way is attached to the structure such that it forms an integral part of the femur. The follower of the cam is connected to a long link which connects the follower to the foot. So when the knee (along with tibia) is rotated, then we have a relative angular rotation between the follower and the cam. The follower and the linkage mechanism produce the required output which is the angle between the foot and the tibia.

ISSUE FACED:

The issue in this case is that for a single orientation of the knee, multiple angles need to be achieved at the ankle. That is, for a given knee flexion in STANCE phase, there is a certain ankle flexion angle to be generated, with reference to the normal gait. For the same knee flexion angle in the SWING phase, the ankle flexion to be achieved will be different. If the structure indicated previously were used, then both the cases of knee flexion would give only one ankle flexion.

Hence, it was understood, that there is a need to obtain a more unique characterization of each stage in the gait cycle. Once such uniqueness is ensured then each stage will correspond to a particular ankle flexion and that can be achieved by the mechanism. The CAM angle must characterize each stage more uniquely. This is only possible if the major portion of CAM action is MONOTONIC. If not we will have similar issues mentioned before. The CAM should be able to rotate through a monotonic angular range for major part of the gait.

RECTIFICATION OF THE ISSUE:

The issue is that monotonicity of the input is not satisfied. To do this we have to convert the rotation of the knee to a single direction motion and then feed it to the cam. Possible ways thought of to do this are

1. Clutch-Spring Mechanism

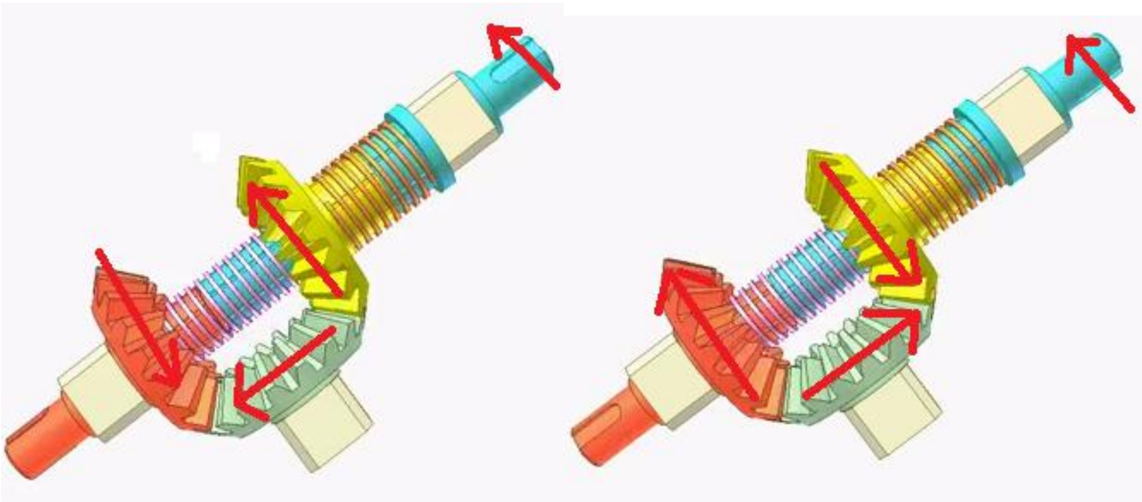


Fig.6 Clutch spring mechanism in which output (blue) rotates in same direction irrespective of direction of rotation of input (red)

2. Gear Mechanism.

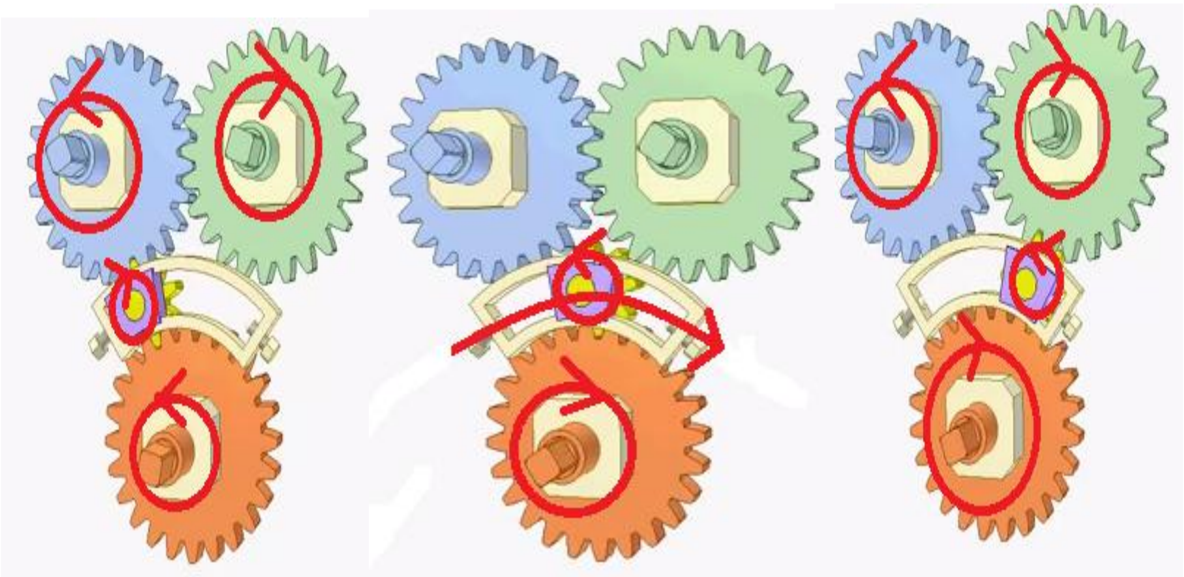


Fig.7 Gear mechanism in which output (blue and green) rotates in same direction irrespective of direction of input (red) but with a lag for the yellow gear to travel between gears

The issue with the above two mechanisms is the housing of the gear box. The housing has to be held in a fixed orientation as well as the housing should be displaced as the knee is displaced in the gait cycle. This solution in that way seems a little absurd. In addition to that, there will be a lag in the gear mechanism which is highly undesirable in our mechanism as the whole gait cycle will go out of phase. There seem to be too many mechanical complications involved.

Another possible way is to take a superposition of two inputs. Another input that we could take is the rotation of the hip angle. Some possible ways of using this are

1. Come up with a mechanical linkage with Double-Input-Single-Output system. But this is highly complex.
2. Use a linear combination of the two inputs to construct a monotonic input. Then the issue of retracing the path will be satisfied. This is achieved by giving freedom to the follower guide-way and connecting another link which joins the follower's guide-way to the hip joint.

FINAL SOLUTION:

- We use a cam which takes input from the knee joint as before.
- The cam is in contact with the follower whose guide-way is attached by a revolute joint to the knee joint unlike the cam which is rigid.
- Here we took the simplest case where the follower is always vertical. To achieve this, a little above the pelvis, we attach a slider and a T bar connects the slider to the follower's guide-way.
- When the knee joint displaces, then the follower is also displaced but it remains vertical because of the T joint but the T joint slides along its guide-way which is above the pelvis. The other end of the follower is connected to the foot by the longest link as before.

The outcome is that the input increases monotonically for about 64% of the gait cycle and in the remaining 36% of the cycle the traced path is just retraced in a faster fashion. As an outcome, during swing phase, the dorsiflexion is higher than normal but the ground is cleared and after Feet Adjacent the ankle plantarflexes than what is required which is again not an issue as the shock is less when loading response occurs.

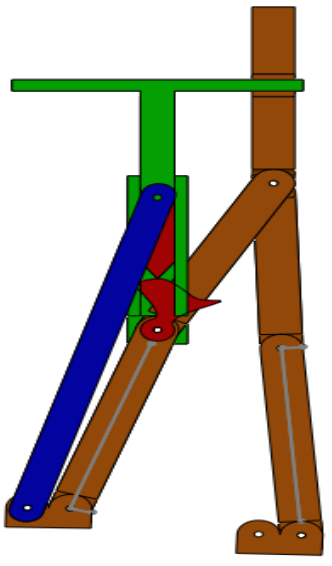


Fig.8 Solid works model of the mechanism

SOLID WORDS MODEL SHOWING DIFFERENT POSITIONS

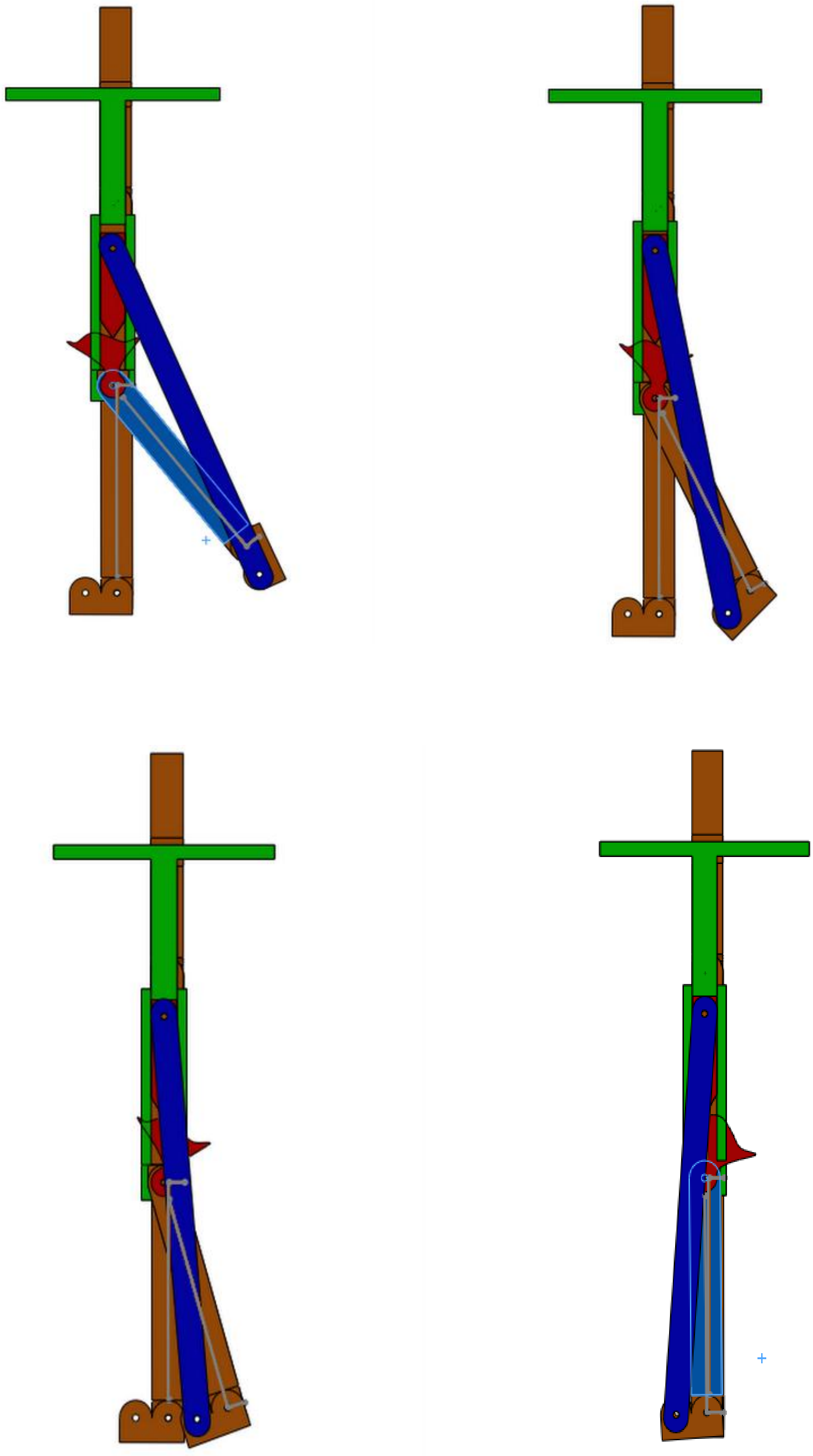


Fig.9 Solid works model showing different positions of the proposed mechanism

ANALYSIS OF THE PROPOSED MECHANISM

ASSUMPTIONS INVOLVED

1. The orthotic mechanism is assumed massless. The forces are not high to give rise to significant stresses. So the material does not have to be strong and heavy. It can be light weight. The weights of the links are neglected.

2. The dynamics of the mechanism to keep the follower vertical is neglected. It just guides the guide-way into position. Suppose you are standing and you flex your hip. Then the guide-way immediately rotates about its hinge joint and adjusts its angle. The follower slides through the guide-way and there is not force on the ‘above-hip’ portion. To be precise, suppose the guide was jammed at the thigh and you lift the thigh, then it will jam with a lot of force against the upper body. But since it has a slider and so it won’t cause much force and slide into place as it is just a guiding mechanism and not a driving mechanism. The forces involved there will be less.

3. The Ground Reaction Force for the orthosis imposed gait is assumed same as that for a normal gait cycle.

4. The moment contribution is assumed to be from the predominant muscle activity

5. External links are also assumed frictionless in addition to the internal joints

6. The horizontal component of the force acted on the follower by the connecting link substituting the tibialis anterior muscle is neglected.

7. Analyses are done only in the sagittal plane.

8. The centre of mass of the body does not oscillate vertically.

9. Segmental weights and centre of gravity are known

10. Effects of tendons and ligaments are neglected

11. Anatomical axes of joint rotations are known

12. Axis of joint rotation does not change.

13. Due to unavailability of the data for the triceps surae moments when the orthosis is added, we use the moment data of the normal gait cycle itself for analyses.

DESIGN OF CAM PROFILE

The following section deals with the design of the CAM profile for the mechanism. The first step is the identification of the input and the output of the CAM. It is important to note that CAM does not actually rotate about the hinge at the knee. It is rigidly connected to the tibia. When the tibia rotates, the CAM makes an angle with the vertical follower. In the reference frame of the centre of the CAM, this motion would result in a regular CAM-follower motion which has been widely studied.

This angle serves as the input to CAM rotation. Geometrically it can be shown that the angle made by the CAM with respect to the follower at each instant of the gait is given by:

$$\text{Input angle} = \text{Knee flexion angle} - \text{Hip flexion angle}$$

This input is plotted for the entire gait cycle. The following plot shows the variation with respect to time.

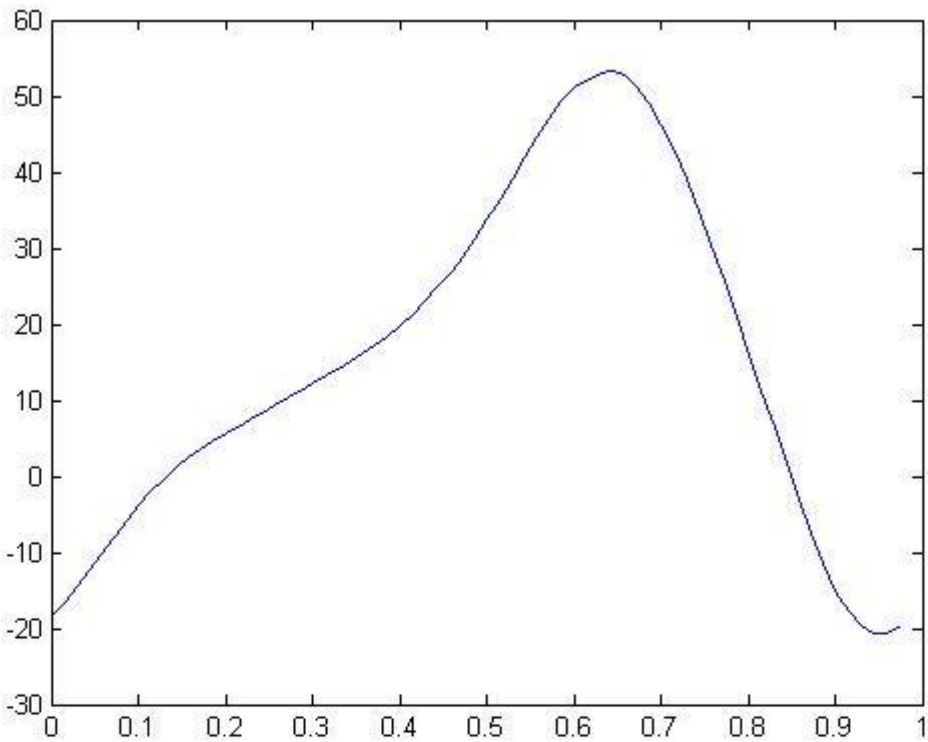


Fig.10 Input angle to the CAM

It can be seen that monotonocity is maintained for more than 60% of the cycle. After that the CAM rocks backwards and the same angles are retraced at a faster pace.

The ankle joint motion in the normal gait cycle is given by the following plot:

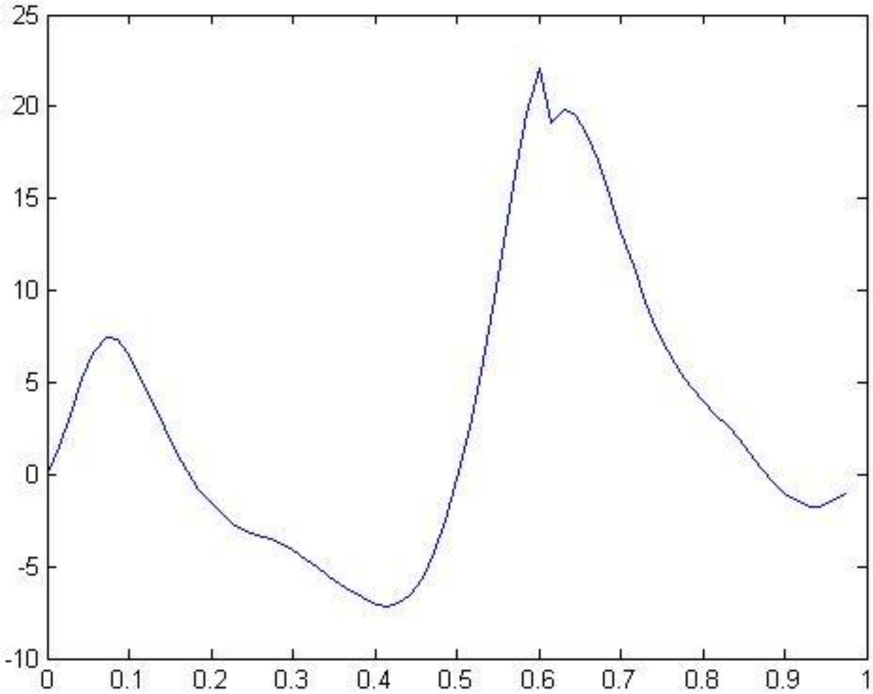


Fig.11 Normal gait ankle joint motion

The small spike during swing is because of a little aberration in the data. Anyway, ideally this is the plot we aim to achieve using the CAM driven mechanism.

From the input angle of the CAM it is clear that the CAM rocks back faster after about 60% of the gait cycle. Suppose we have designed the CAM profile such that it can cause the ankle to trace the exact profile as the normal gait, in one rock(during monotonocity) of the CAM. When it rocks back, it will make the ankle to re-trace the same path which it would have taken till that point. Hence, if the ankle was driven by this mechanism, then the following plot is obtained(the 'red' plot):

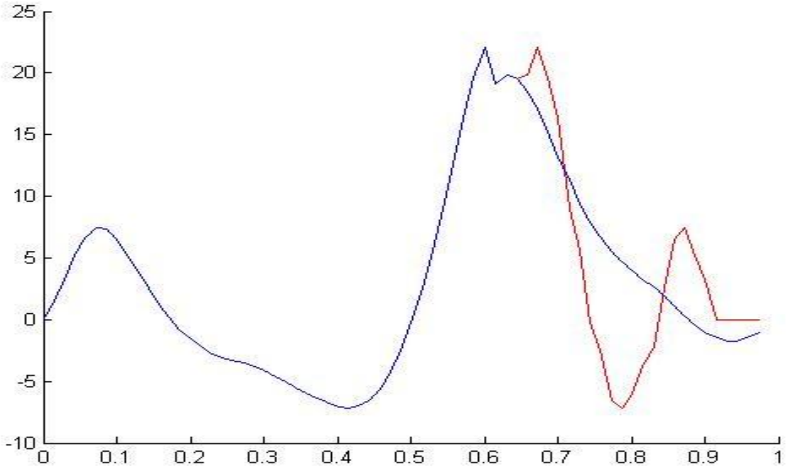


Fig.12 Fully CAM driven ankle vs. normal gait

We observe that there is excess dorsiflexion followed by excess plantarflexion occurring at the end. As a lucky consequence, the plantarflexion happens much after 'FEET ADJACENT' and hence the foot safely clears the ground. But the excess dorsiflexion is an issue because it might cause excess strain on the Triceps surae. So, we intend to reduce this part.

The excess dorsiflexed part occurs because of the existence of the ankle rocker region to the left side of the peak. That part is being reflected. So, reducing the excess dorsiflexion implies the flattening of the dorsiflexion of the ankle rocker phase. This is **not desirable** because ankle rocker is a important stage in the gait cycle. Reducing the extent of dorsiflexion in the phase will cause discomfort and reduce step length.

An interesting solution was devised. In the ankle rocker phase, the foot is flat on the ground. Hence it is bearing against the ground. By **bearing against the ground** the person can easily dorsiflex his ankle without the need of the linkage. The CAM need not drive the ankle in this region. Hence, in this region we allow the GRF to help the person dorsilfex his ankle.

This gives the freedom to tamper with the CAM-driven profile of the region. The statement sounds very ambiguous. What is implied is that in the ankle-rocking phase, the person uses the GRF. The CAM mechanism is not required. The CAM need not dorsiflex the ankle in that phase. It can take over, once the plantarflexion begins just at the end of ankle-rocker phase.

So, the ankle is initially CAM driven, then GRF driven, and then CAM driven again. The following plots bring out the distinction:

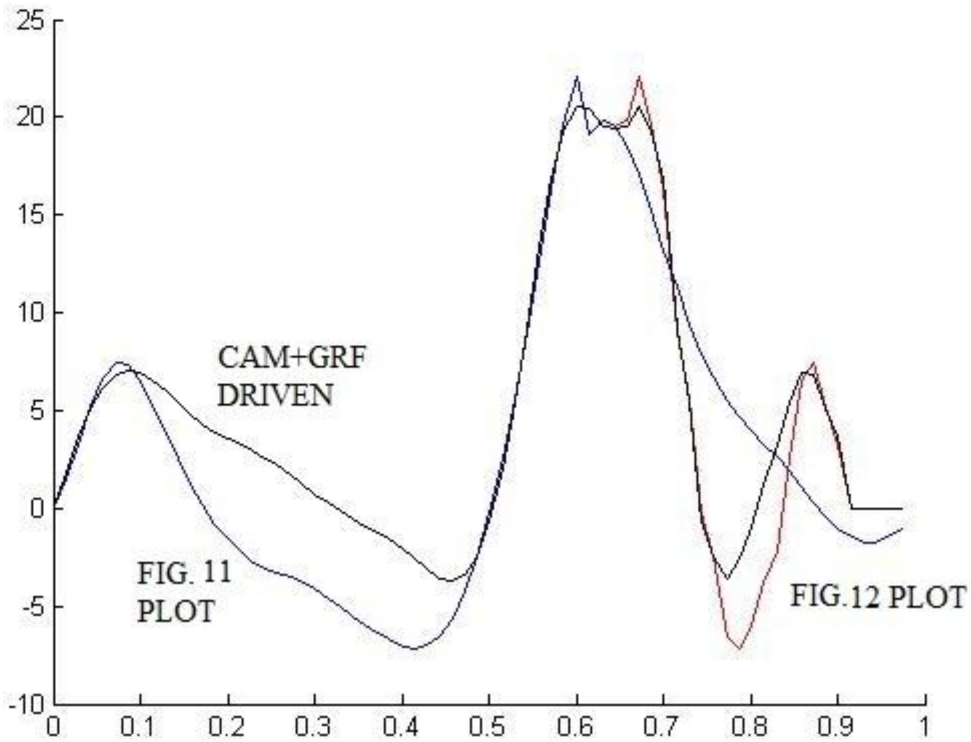


Fig.13 CAM+GRF driven ankle vs. previous profiles

There are three plots. Comparing to the first ankle joint motion plot, there is one plot which goes through the normal gait motion. Then, the second plot of the fully cam driven ankle is shown. Then the reduction of the excess dorsiflexion is also shown by the third plot.

If the person just hung his foot in the air, and the knee and the hip went through the GAIT motions, then the path traced by the ankle would be like the third plot i.e. the CAM+GRF driven plot. But due to presence of the GRF, the ankle rocker phase is made to dorsiflex more than what the CAM stipulates. In this region the follower detaches from the CAM, and the CAM no longer drives the ankle.

After the end of the ankle rocker phase, the CAM regains control. The third plot joins the initial plots and then the remaining path is traced with a reduced dorsiflexion in the end. So, the joint motion executed by the ankle is given by:

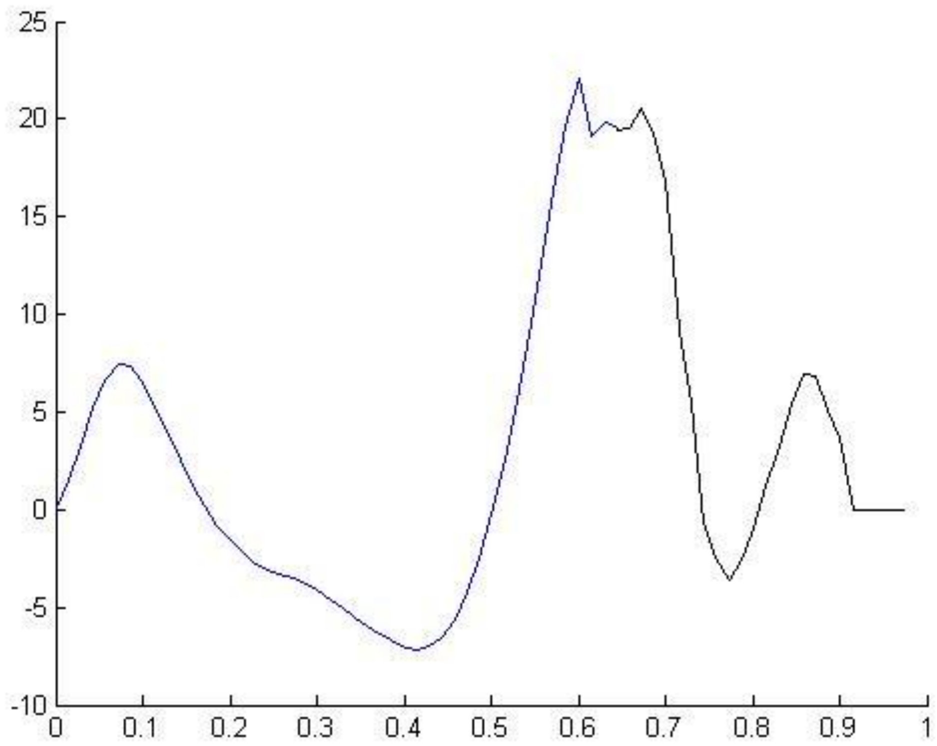


Fig.14: Final joint motion achieved at the ankle

Now, it all boils down to the construction of the CAM. Only the monotonic region of the input is considered. In that region, the input is given by:

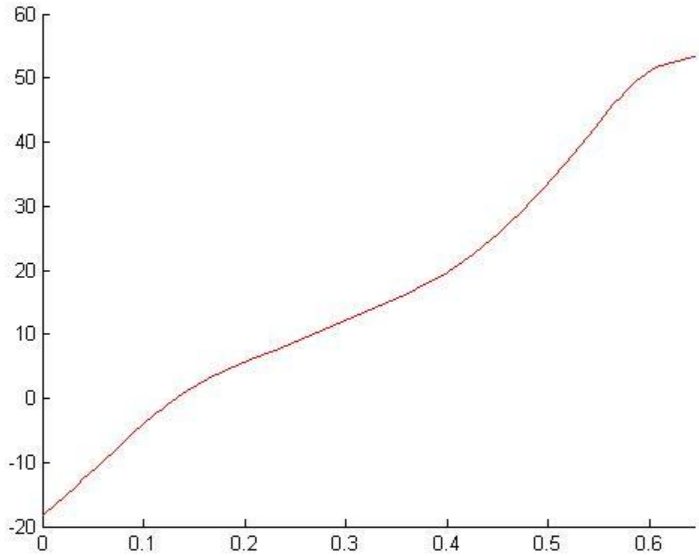


Fig.15 Input to CAM

In order to get the follower output, the mechanism must be used. At each stage, a kinematic position analysis needs to be carries out to determine the position of the end of the follower given the position of the ankle. The ankle motion follows the CAM driven profile mentioned before. Using that, and performing kinematic analysis using a MATLAB code, the following output is obtained at the follower:

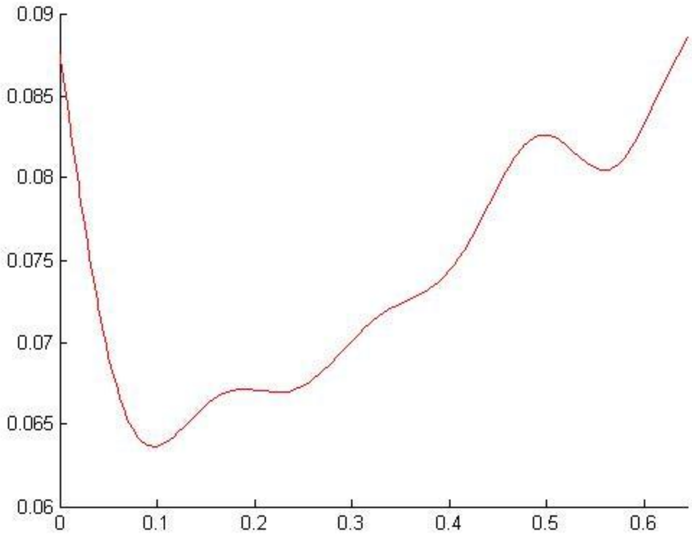


Fig.16 Follower output

Using the input angle and the follower output, and using a knife-edge follower, the following CAM profile was generated.

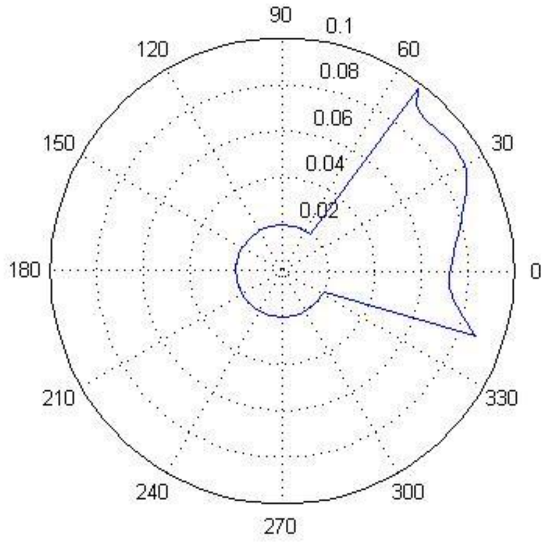


Fig.17 CAM profile

It is important to note that the CAM has the required profile only for one sector. It rocks within this region and hence, the other parts do not need to have a specific profile. This CAM profile was incorporated into the SOLIDWORKS model of the mechanism.

Now that the mechanism is ready, the next section deals with the force analysis of this mechanism.

FORCE ANALYSIS

This section shows the force analysis that is done to validate the feasibility of the proposed mechanism.

Basically we were looking to evaluate the forces and moments at each instant of the gait cycle with and without the orthosis.

For the above purpose, we assumed a linkage model of the lower limb of the human body. The linkage model assumes the lower limb to be made of 3 rigid segments, i.e the thigh, shank and the foot are modeled as rigid entities connected to each other by revolute (pin) joints. This enables the free rotation of each rigid link about their respective revolute joints.

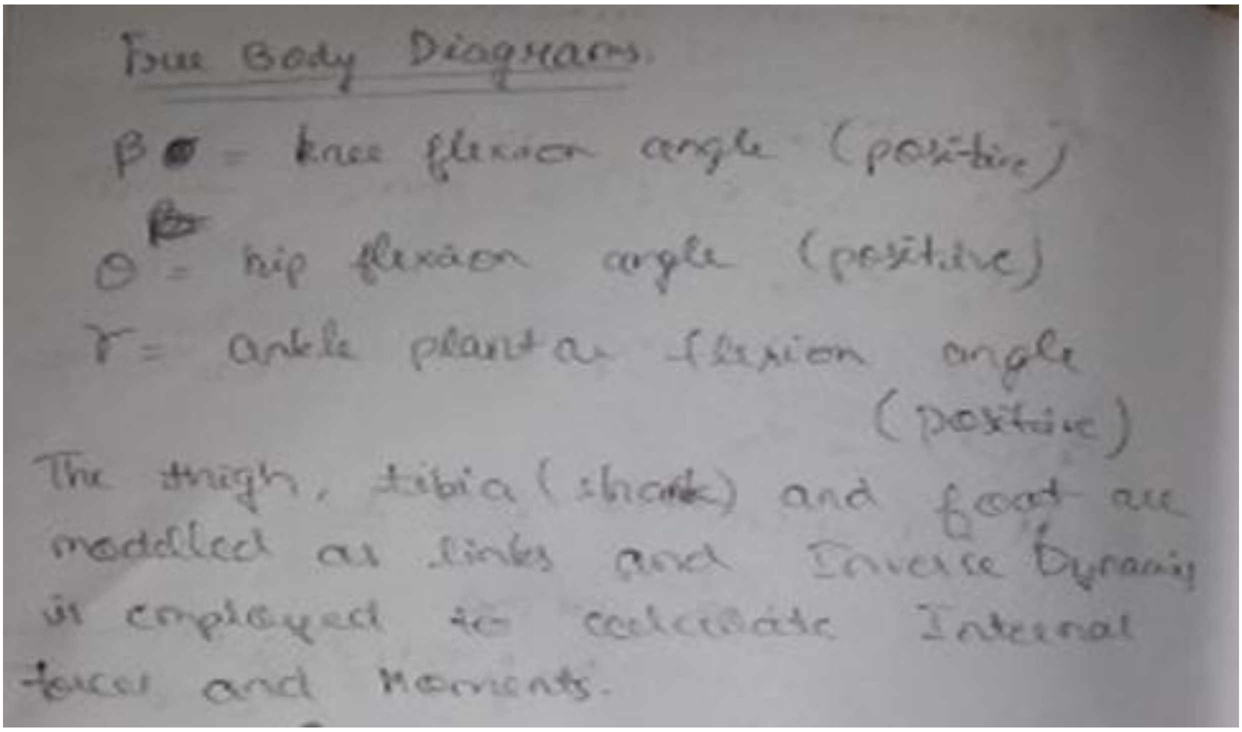
At each joint, there is a joint reaction force and a net moment due to various muscles acting in the vicinity of the joint. These are the unknowns that we must determine. Note that the analysis is purely 2D in nature, and hence for each segment, there are 3 unknowns and 3 equations resulting in an exact solution. Here, we are assuming that the kinematic details of each link (linear accelerations, ground reactions, angular accelerations etc.) are available to us i.e these parameters are known.

Such an analysis of determining the dynamics (internal forces and moments)of the system from the kinematics is called “Inverse Dynamics”.

We first tried to replicate the normal gait using our linkage model of the lower limb. This was done to validate our assumption of a link model.

The free body diagram of the foot is solved first and then the shank. We are primarily concerned with the ankle and the knee joint and hence, the analysis of the hip joint was left out. The reasons for this will be clearer soon.

Following are the FBD diagrams of the foot and the knee.



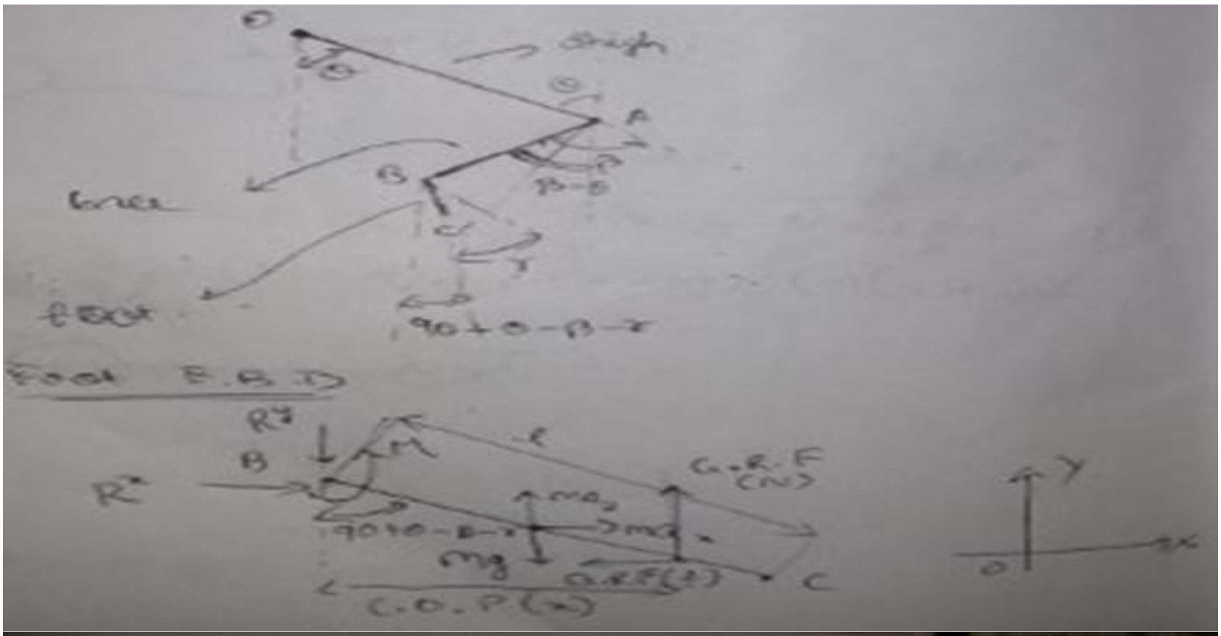
Free Body Diagrams.

β = knee flexion angle (positive)

θ = hip flexion angle (positive)

γ = ankle plantar flexion angle (positive)

The thigh, tibia (shank) and foot are modelled as links and Inverse Dynamics is employed to calculate Internal forces and moments.



3 equation and 3 unknown
(R_x, R_y, M)

$\Sigma F_x = ma_x$

$\Rightarrow R^x - G.R.F(t) = ma_x$ —— (1)

$\Sigma F_y = ma_y$

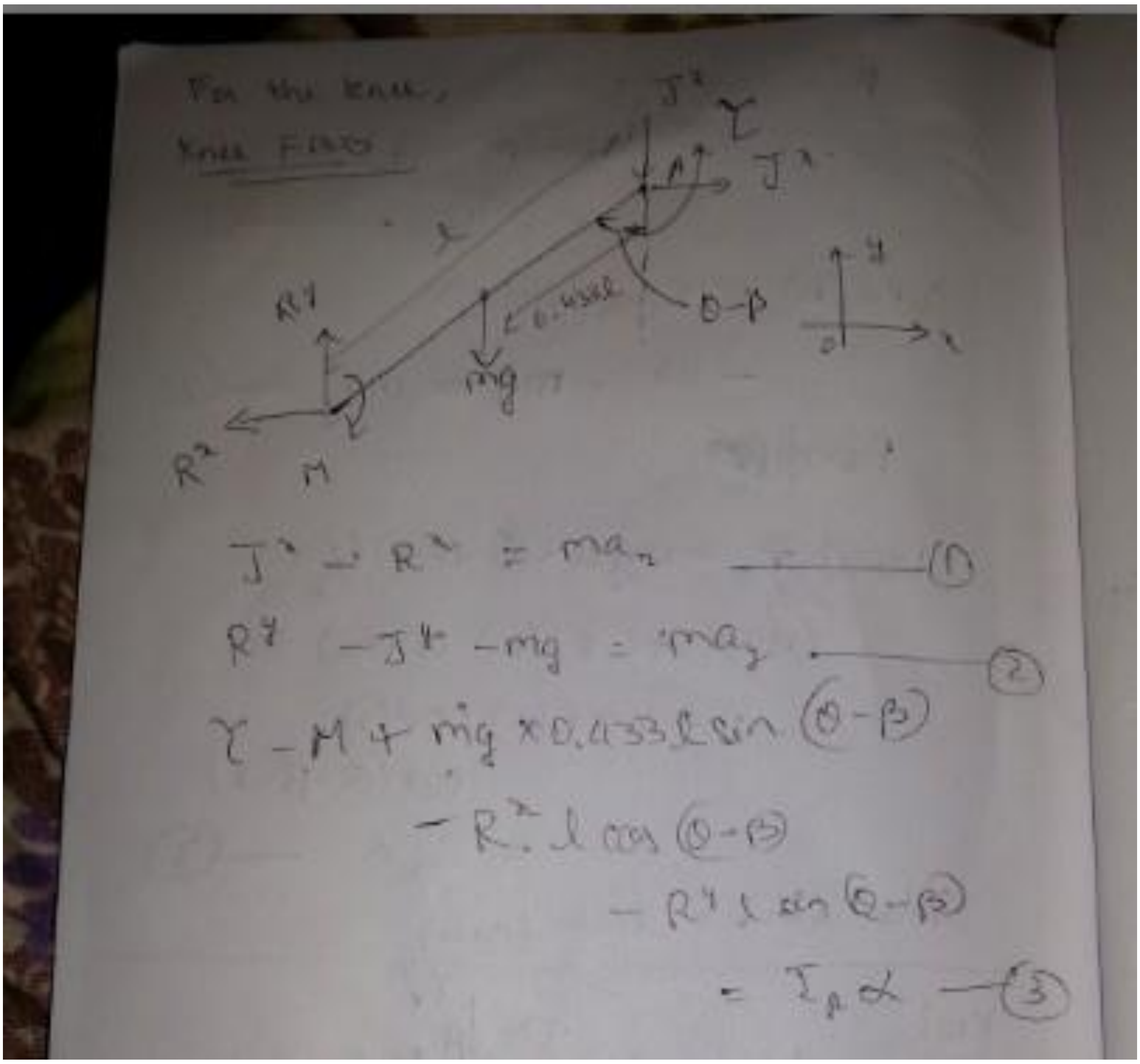
$\Rightarrow G.R.F(N) - mg - R^y = ma_y$ —— (2)

$\Sigma M = I\alpha$

$\Rightarrow M - mg \sin(\theta - \beta - \gamma)\frac{l}{2} + G.R.F(N) \cdot COP(x) + G.R.F(t)\cot(\theta - \beta - \gamma)\, COP(x) = I_B\,\alpha$ —— (3)

$\Sigma M = I\alpha$

$M - mg \sin(\theta - \beta - \gamma)\frac{l}{2} + G.R.F(N) \cdot COP(x) + G.R.F(t) \cdot COP(x) \cdot \cot(\theta - \beta - \gamma) = I_B\alpha$ —— (3)



The above diagrams were for the normal gait.

We used the expressions derived to do run a simulation on MATLAB where the Forces and Moments were determined at each instant and the results were compared with the experimentally obtained forces and moments. The following graphs illustrate the point.

Plots of the ankle forces and moments

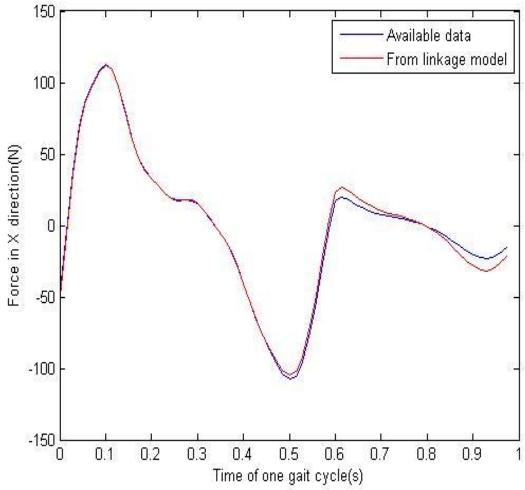


Fig.18 X direction force in ankle joint

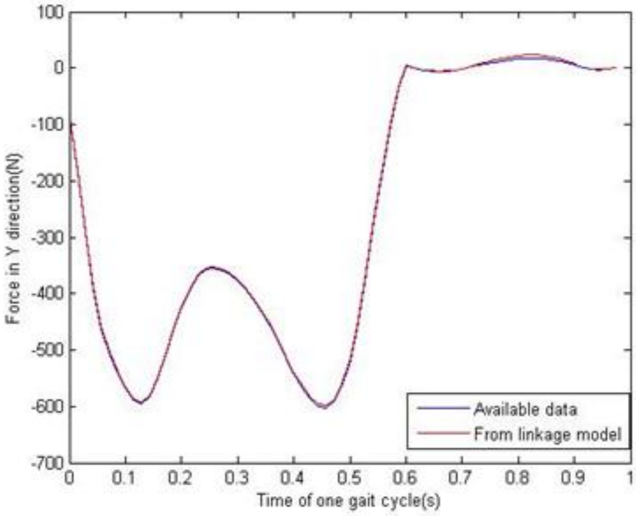


Fig.19 Y direction force in ankle joint

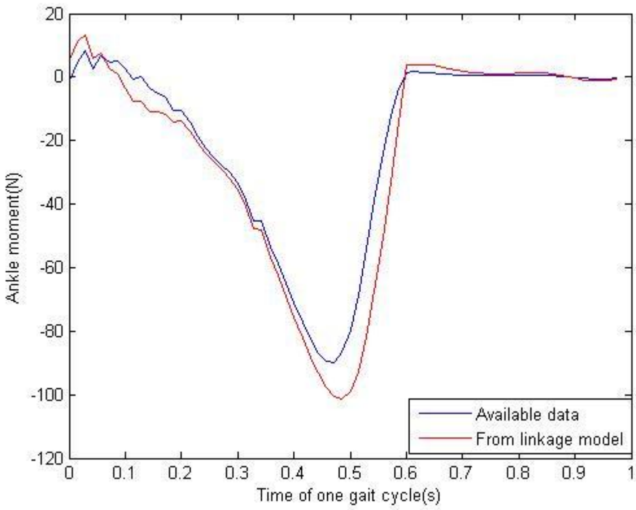


Fig.20Moment in ankle joint

Plot of the knee forces and moments

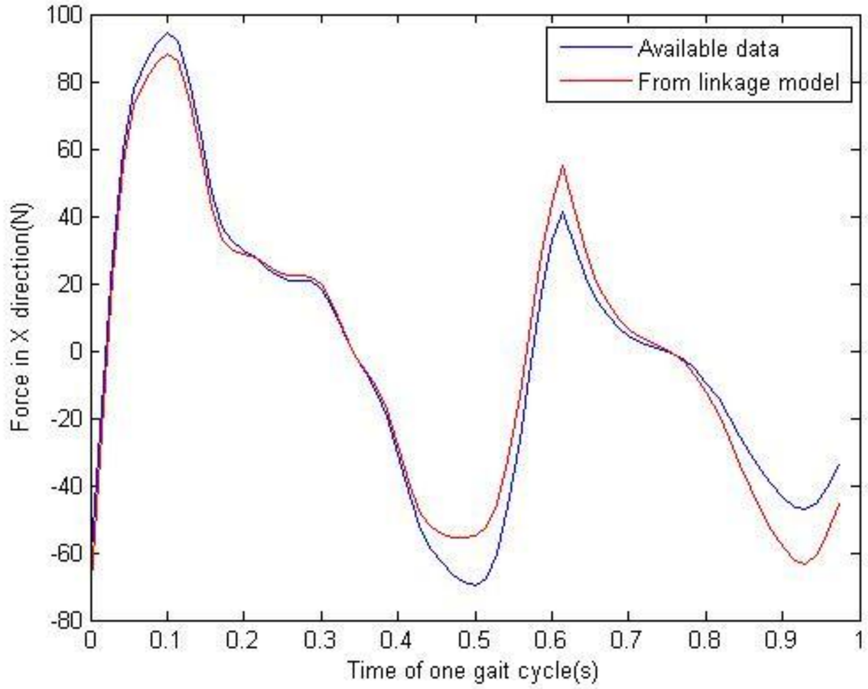


Fig.21 X direction force in knee joint

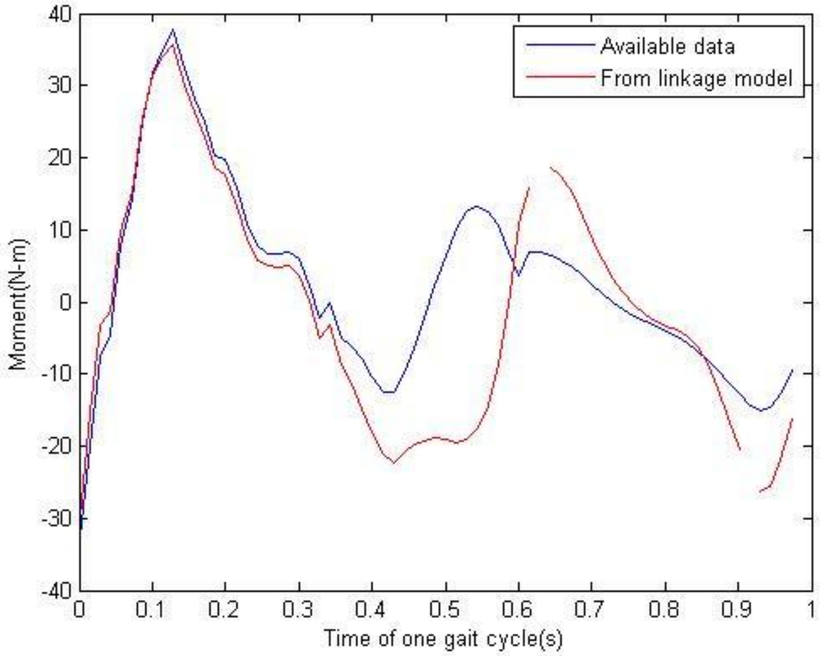


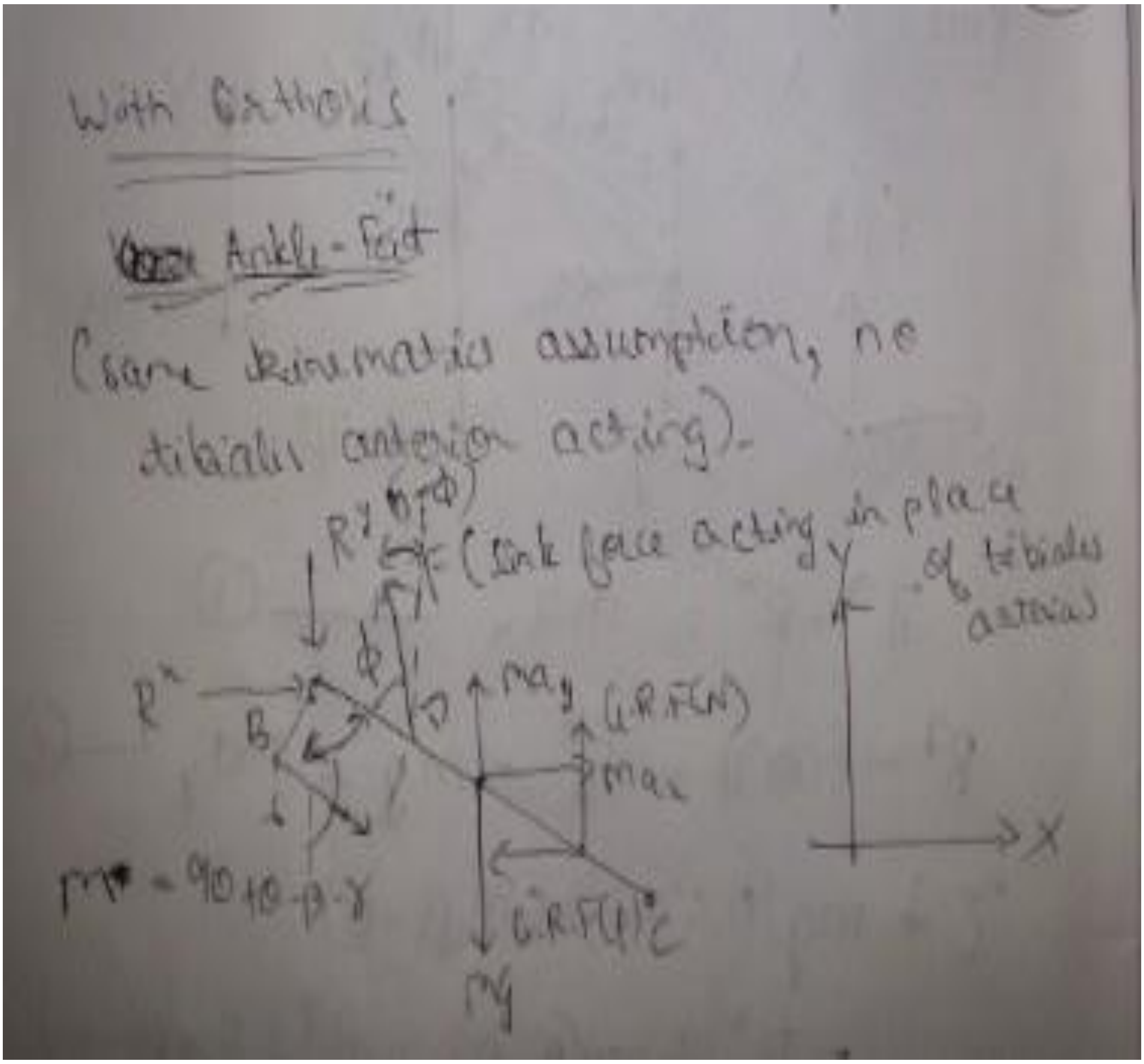
Fig.22 Moment in knee joint

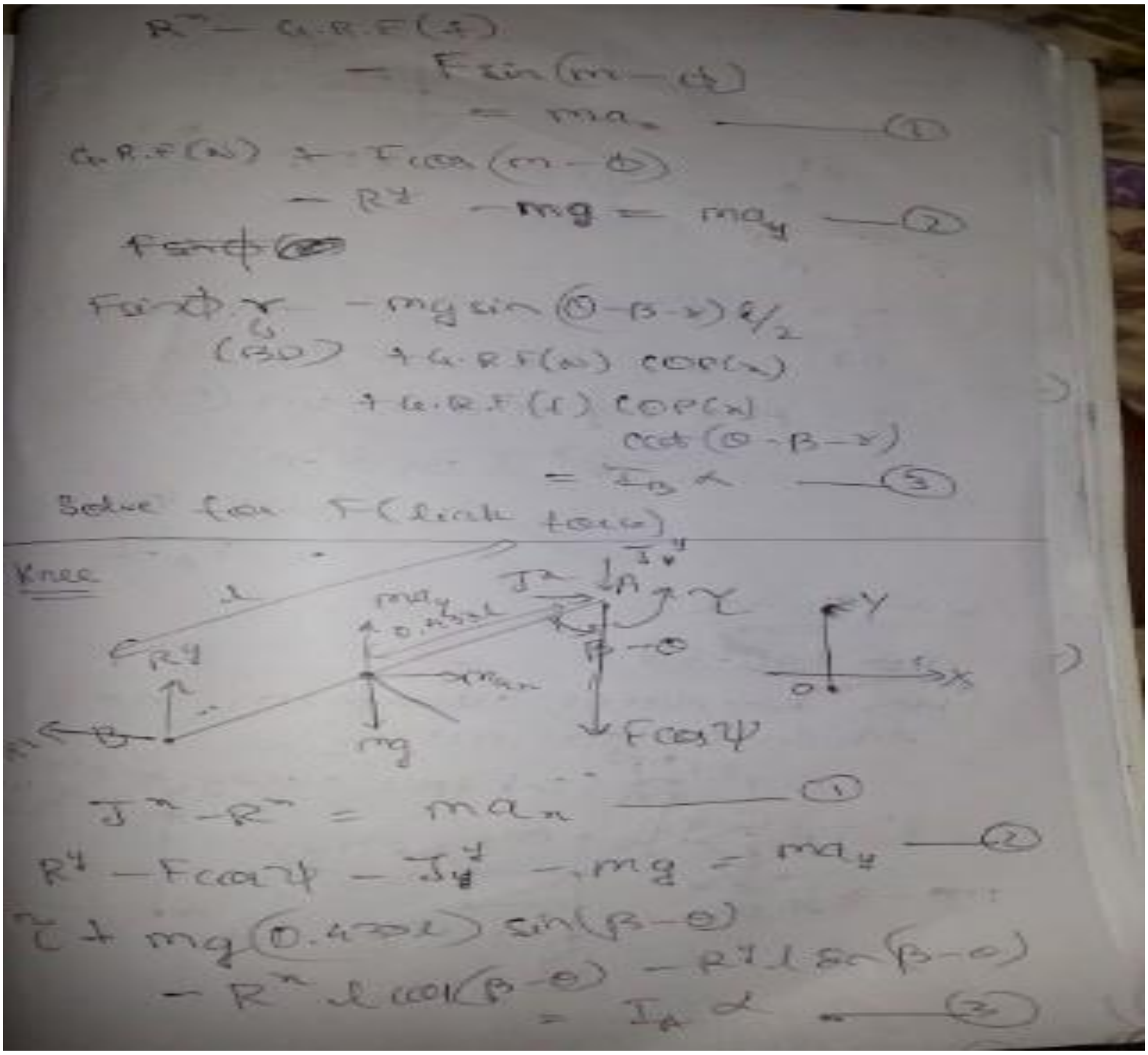
As can be seen, the forces and moments are reasonable validated and hence we can conclude that our linkage model to be a good approximation of the actual dynamics. We will proceed to calculate the same forces and moments with the orthosis attached to the pathological gait.

Note that the orthosis is designed to replicate the normal gait in theory. We will therefore assume that with the orthosis, the gait kinematics remain unchanged. We only assume that the tibialis anterior no longer acts (absent) and the link force compensates for the same.

We have additionally assumed that the weights of the CAM and the linkages are negligible with respect to the weights of the thigh, shank and the foot.

The Free body diagrams with the orthosis are shown below.





With the above equations of motion, we ran simulations in MATLAB to get the forces and moments at the ankle and knee joints. These forces and moments are compared to that obtained in the normal gait cycle .

The orthosis basically substitutes for the absence of the tibialis anterior i.e it helps in stabilizing the foot eccentrically during heel contact and subsequent heel rocker motion, and the concentric dorsiflexion of the foot required during pre-swing for clearance from the ground.

We have not neglected the action of the triceps surae in the pathological gait and hence, the orthosis does not act whenever the triceps surae is the dominant muscle in the gait cycle.

The orthosis essentially uses the knee motion to produce ankle motion. As a result, there is an increased load on the quadriceps acting on the knee. We hope to show that the additional load is not too large i.e. it is not a burden for the person during his/her gait cycle.

Plot of the ankle forces

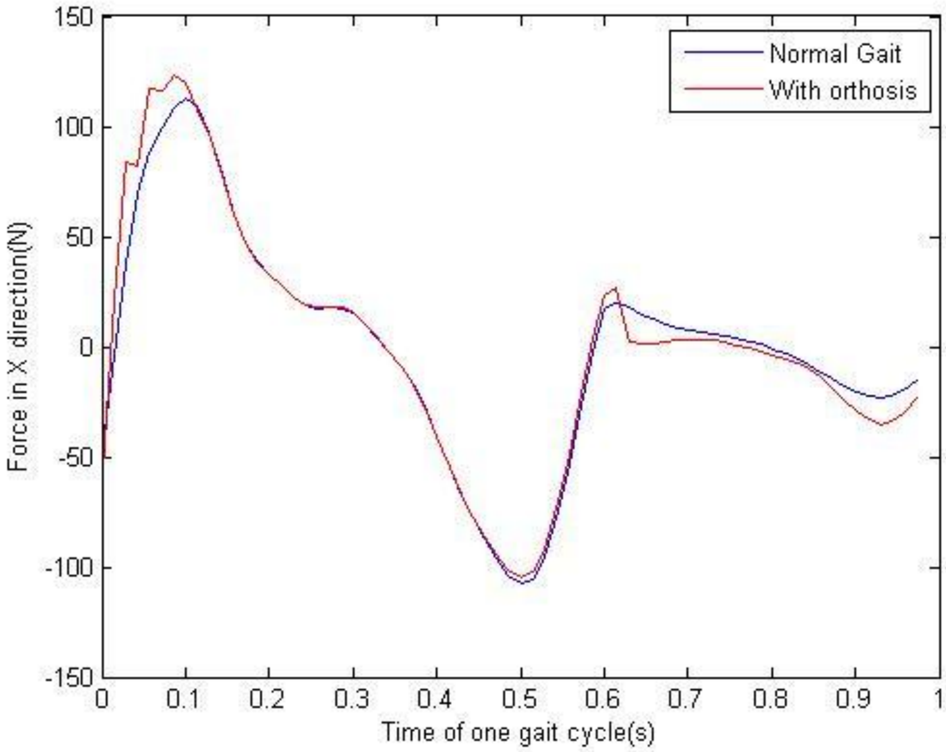


Fig.23 Force in X direction in ankle joint

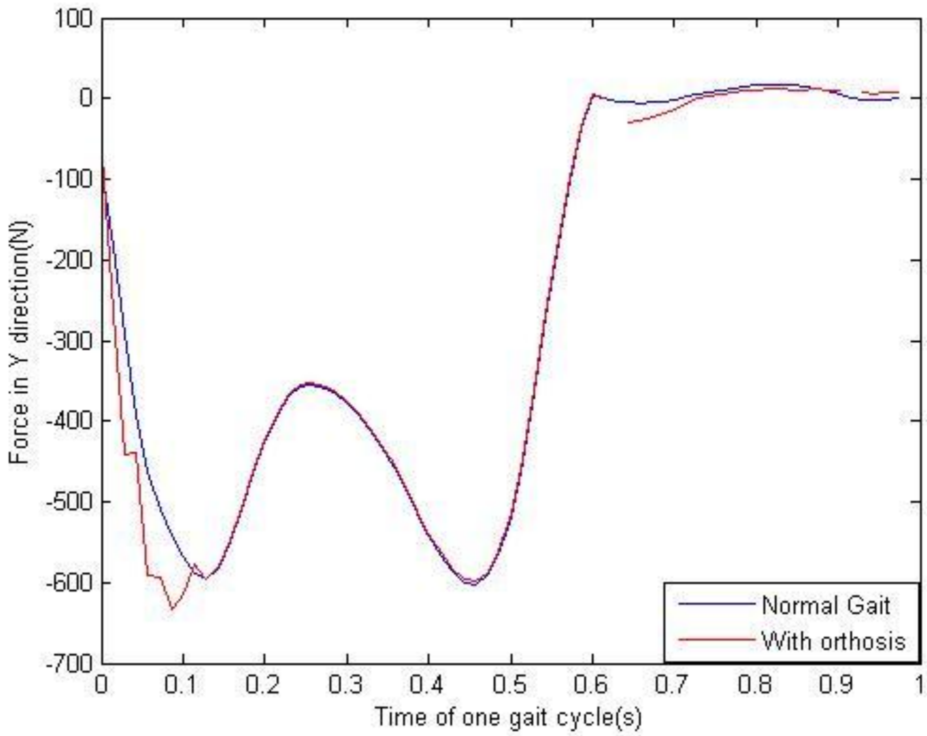


Fig.24 Force in Y direction in ankle joint

Plot of the knee forces and moments

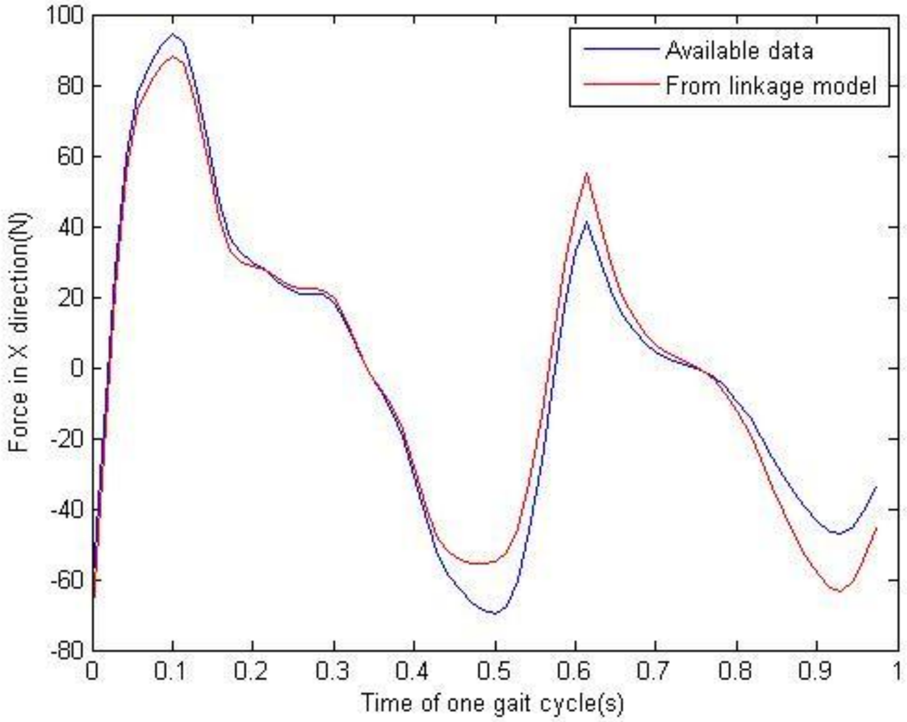


Fig.25 Force in X direction in knee joint

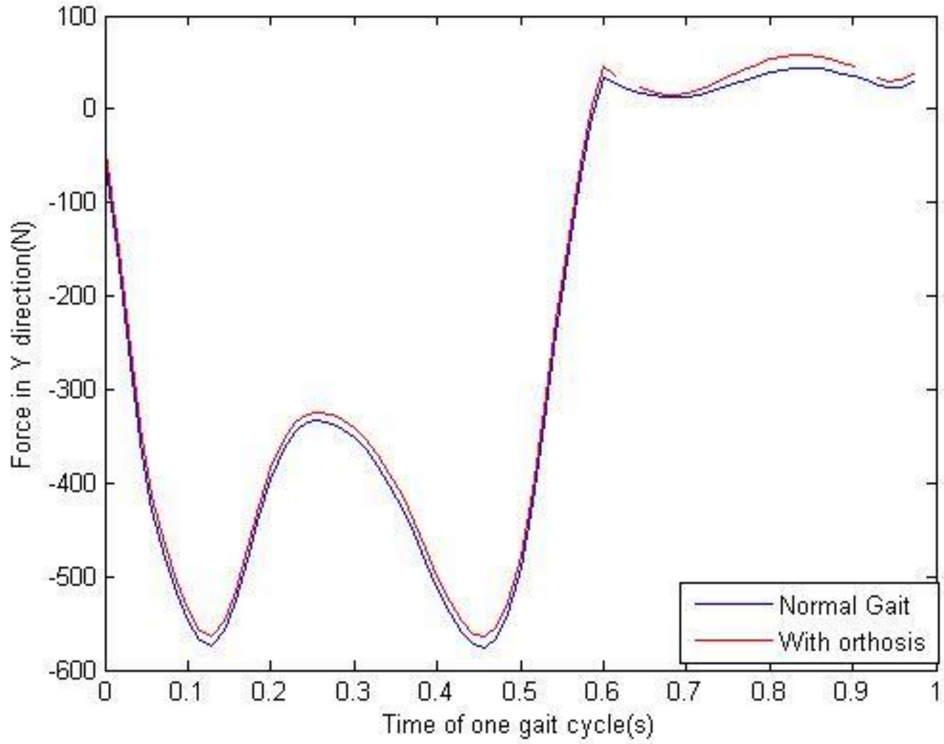


Fig.26 Force in Y direction in knee joint

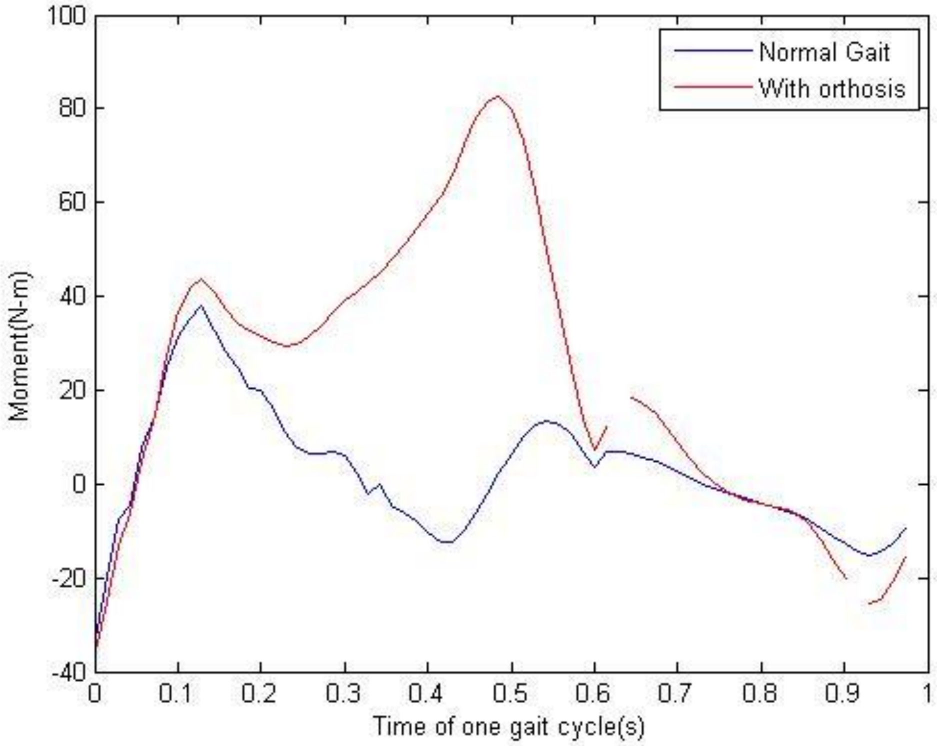


Fig.27 Moment in knee joint

The plot above is the most crucial one. We can see the moment generated in the knee because of the orthosis. The quadriceps are indeed responsible for control of flexion/extension at the ankle in the absence of the tibialiis anterior.

The moment exerted by the quadriceps increases to a peak of 80 Nm at about 0.5 seconds into the gait cycle. Apart from this peak, the deviations are fairly reasonable. The 80 Nm moment is also not too high. A pretty strong set of quadriceps can withstand the additional moment.

The joint reactions are not affected much by the orthosis as can be observed. There are slight deviations (increased forces) during the heel rocker and the pre-swing phase.

We therefore conclude that the Moments are internal forces with the orthosis are not too deviant from the moments and the forces in the normal gait cycle.

The mechanism can be developed and analysed further with better assumptions to get a better estimate.

CONCLUSIONS

We see that the mechanism hold goods for our first basic analysis. Later it has to be tested by including the effects of the mass and the effects of the T bar mechanism. Some drawbacks of the mechanism are:

1. The T bar mechanism can cause some hindrance.

2. The perfect gait cycle cannot be achieved.

3. The mechanism is not suited for other actions like squating

Yet, the mechanism can do a good job in the performance with respect to the gait cycle. Further improvements can be to bring in electrical components to replace the mechanical components but it is a trade off between the complexity of the mechanism and the failure and maintenance of electrical components.

REFERENCES

1. Gait Analysis-Normal and Pathological by Jacquelin Perry M D

2. Gait Analysis and Angular Kinematics by Winter

3. http://emedicine.medscape.com/article/320160-overview

4. https://www.youtube.com/watch?v=AQRuCse7ENY

5. https://www.youtube.com/watch?v=esVq6jfTigM&list=PL1VUU5SuOubn3hxhoXrB-

2i2cw0oiCkZh

6. http://www.rehab.research.va.gov/jour/11/484/page459.html

7. http://en.wikipedia.org/wiki/Foot_drop

8. https://www.youtube.com/watch?v=SWvEU8FYMFc

9. https://www.youtube.com/watch?v=8c4bGhvK0Qs--