DEVELOPMENT OF 3D GAIT ANALYZER SOFTWARE BASED ON MARKER POSITION DATA

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Abstract

Human walking analysis is instrumental in medical rehabilitation because it provides quantitative information of human body segment during walking. The present work is a part of a research on the development of 3D gait analyzer software. The software consists of a program to process and display 3D kinematics and kinetics parameters of human gait based on position of external skin markers as a function of time in X, Y, Z axis and anthropometric data of a subject. To achieve good results, the marker position data is initially smoothed to eliminate noises before further processing. A graphical user interface is also constructed in this work for ease of use of the program and to enable presenting the results readily. The program developed in the present research could successfully calculate parameters of human gait such as spatio temporal parameters, linear kinematics, angular kinematics of joints, as well as ground reaction and joint forces.

Keywords: 3D kinematics, 3D kinetics, Gait analysis, Human body model, Inverse dynamics

Introduction

Human movement analysis has long been studied and applied in clinical applications. The one that specifically study the characteristics of human walking is called gait analysis. Gait analysis is used by clinician(s) to assist in the selection of the most appropriate treatment for the patient. This is based on one or more of: *diagnosis* between disease entities, *assessment* of the severity, *monitoring* progress in the presence or absence of intervention, and *prediction* of the outcome of intervention [1].

In general, human gait analysis can be measured by direct measurement techniques and also imaging (optical) measurement techniques [2]. Examples of direct measurement are the use of goniometer and accelerometer. However, each device has disadvantages. Goniometer device is difficult to fit and constrains the movement of the subject. The output is unreliable and it only measures relative angular data, not absolute values. For the accelerometer, its signal depends on the position of the accelerometer on the limb. Moreover, the wiring of the accelerometers can also constrain the movement of subject [3].

Most of the problems encountered by direct measurement techniques could be overcame by imaging (optical) measurement techniques. By using this technique, the position of skin-markers trajectory during walking could be obtained. These data could be further processed to obtain desired kinematics and kinetics parameters of human gait.

Recently, the authors have developed an affordable system for 2D kinematics and kinetics analysis of human gait [4] by using a 25 *fps* home video recorder. The system is further improved to overcome the occlusion problem of markers [5] and has been successfully used to determine *2D* gait parameters of Indonesian people as an effort to develop the first Indonesian gait database [6], [7]. However, information obtained from *2D* measurement, i.e. in sagittal plane, is not as much as information obtained from *3D* measurement. Although the sagittal plane is probably the most important one, where much of the movement parameters could be observed, there are certain gait pathologies where another plane (e.g., the frontal plane) would yield useful information [8]. Therefore, development of a *3D* gait analysis system is needed to obtain gait parameters in frontal and transverse planes.

The present work is a further research aiming to develop software for *3D* gait analysis. Here, a computer program to process and display *3D* kinematics and kinetics parameter of human walking based on position and displacement of external skin markers attached to lower limb of human body is developed. Kinetics calculation is limited to ground reaction forces and joint forces. For the present analysis, dummy data are used [9], in order to check the reliability of the developed program by comparing calculated parameters with those available from literature.

Human Body Model

To observe the kinematics and kinetics of human walking, a model of human body should be constructed first. So far, various *3D* human body models have been proposed in many studies, starting with the simple one by only 5 segments [10], then 7 segments [8], and to the complex ones by dividing human body into 16 [11] and 17 segments [12]. In this research, to avoid complexity, human body is modeled by 8 substantial segments (Figure 1a) as follows: right foot, left foot, right calf, left calf, right thigh, left thigh, pelvic and upper body, which is represented by one segment called HAT (head, arm, and trunk). All segments are linked as one system through 7 joints: right ankle, left ankle, right knee, left knee, right hip, left hip, and lumbar joint as depicted in Figure 1b (blue dot). All joints are modeled as a ball and socket joint which has 3 rotational degree of freedom.

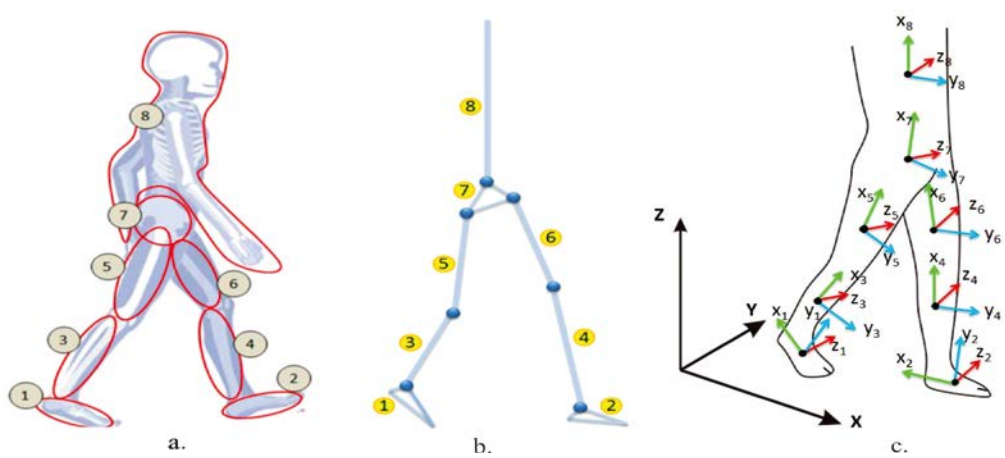


Figure 1. a) 8 segments human body model b) Skeleton model of human body consists of 8 rigid bodies and 7 joints (blue dot) c) Reference axes system

For *3D* analysis, a coordinate system of reference is required, where the axes provide the position and orientation of each segment. Reference axes system described here is based mainly on [8] with addition of reference axis for HAT segment as depicted in Figure 1c. The global reference system (GRS) has X-Y-Z axes, and this axes are fixed for any given camera arrangement. The second reference is a local reference system (LRS) with x-y-z axes, where each of LRS is attached to one segment. The LRS is called anatomical axes and there are 8 anatomical axes to describe orientation of the 8 segments (Figure 1c).

Marker System

The marker system described here referred to Vaughan Kit Marker Set [8]. The set uses 15 markers attached to lower part of human body. The positions of markers are depicted in Figure 2a.

To check whether or not the dummy data that represent the markers position met the system of Vaughan Kit Marker Set, at first, those data is visualized by the developed program as skeleton display as depicted in Figure 2b. Red circles describe positions of markers and the blue ones describe approximate position of the joints.

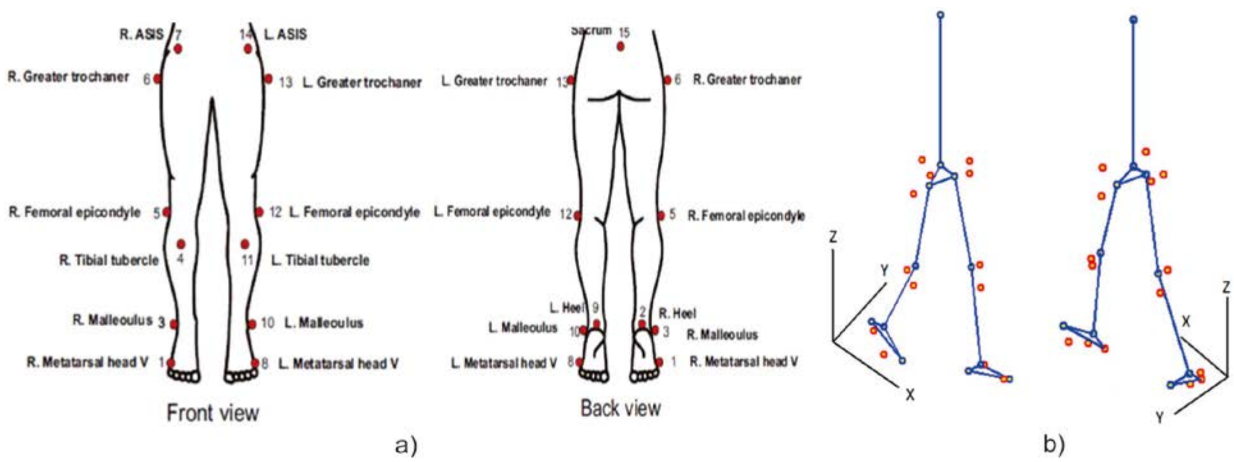


Figure 2. a) Vaughan kit marker set. b) Skeleton display based on marker position data

Smoothing Process

To obtain good results, at first the marker position data should be smoothed to eliminate noises. This step is necessary since calculation of second or third orders parameters such velocity and acceleration is very sensitive to noises. If those data are used directly in the kinematics analysis, the results will be highly inaccurate, as shown by the dashed line in Figure 3a.

In order to remove the noises, smoothing spline technique is performed on the raw data [13]. By using the smoothed data, for example, the foot accelerations are calculated as represented by solid lines in Figure 3a.

To obtain the most appropriate smoothing parameter value, trial and error process is conducted until the obtained curve is close to the one available in the literature. In the process, the acceleration curve of Vaughan [8] is used as reference. The process is illustrated in Figure 3b. Curves on the left are calculation results obtained by varying the smoothing parameter value. The curve on the right is from literature.

The smoothing processes for the kinematic analysis are indicated in the flowchart depicted in Figure 4.

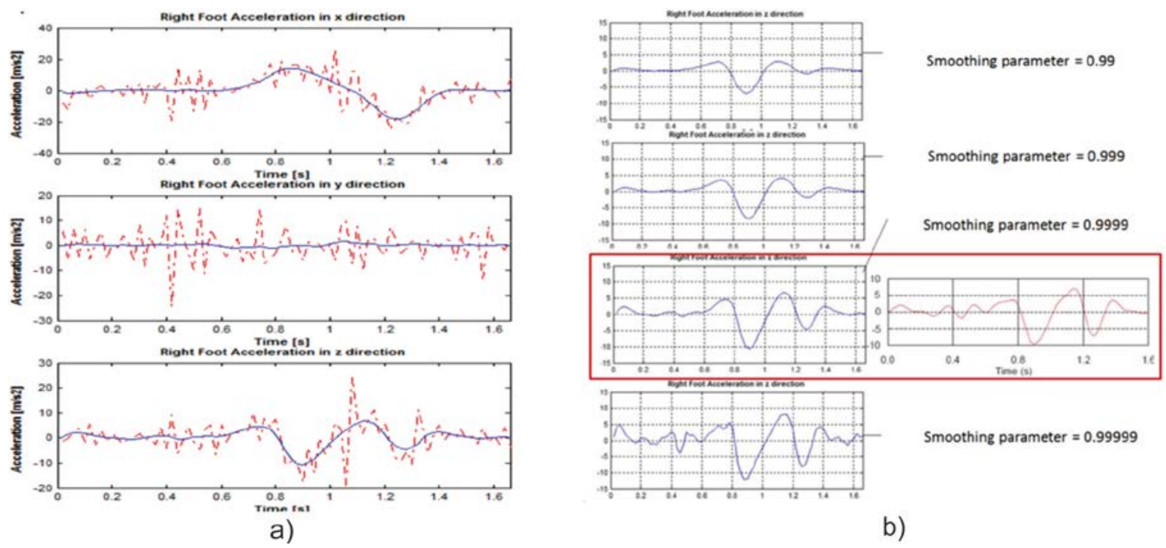


Figure 3. a) Right foot acceleration in x-y-z direction before (dashed line) and after (solid line) smoothing. b) Smoothing parameter selection process

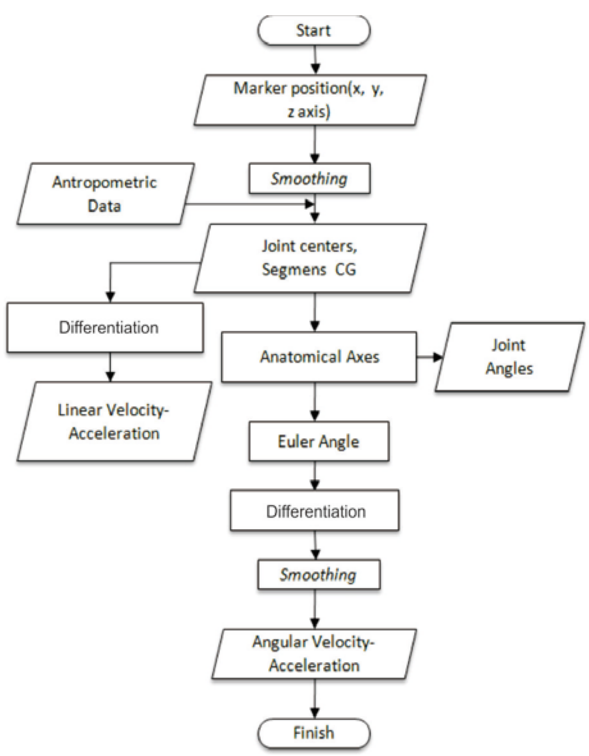


Figure 4. Flowchart of kinematic analysis

Kinetics

Kinetics calculation discussed here is limited to ground reaction and joint forces. To calculate ground reaction forces (GRFs), the mass of each segment must be calculated first. The anthropometric data obtained are utilized to calculate mass of thigh, calf, and foot [8] while pelvic is considered to have 11.7 % of body mass [15]. To obtain mass of HAT, total body mass is then subtracted by the mass of the 7 segments of the lower body. Center of mass (CoM) from thigh, calf, and foot segments are also calculated following Vaughan, et al., (1992), meanwhile CoM of HAT and pelvic are assumed to be in the middle of segments.

GRFs during walking (Ren, et al., 2008) are determined by

$$\vec{F}_{gr} + \vec{F}_{gl} = \sum_{i=1}^{n} [m_i(\vec{a}_i - \vec{g})] \tag{1}$$

where $\vec{F}_{gr}$ and $\vec{F}_{gl}$ are ground forces of right and left foot respectively, m_i is mass of segment i, $\vec{a}_i$ is translational acceleration vector of segment i and $\vec{g}$ is gravitational vector. In case of single support, ground forces on supporting foot can be directly calculated from Equation (1) where ground forces of swinging foot are zero. However, in case of double support the GRFs calculation becomes indeterminate.

To overcome the problem, several methods have been proposed such as ground reaction transfer assumption [16], optimization approaches [17] and smooth transition assumption (STA) [18]. The latest one is adopted for this work.

In general, walking phase could be divided into 4 conditions: right single support, double support with right trailing foot, left single support, and double support with left trailing foot. In case of double support, GRFs on trailing foot are estimated by STA assumption while GRFs on leading foot are calculated from Equation (1).

Center of pressure (*CoP*) during stance phase is assumed to move linearly from heel position (heel strike) to toe position (toe off). *CoP* is defined by

$$CoP_{(i)} = P\ heel_{(t)} + t \cdot \left(\frac{P\ toe_{(t)} - P\ heel_{(t)}}{Ts} \right) \tag{2}$$

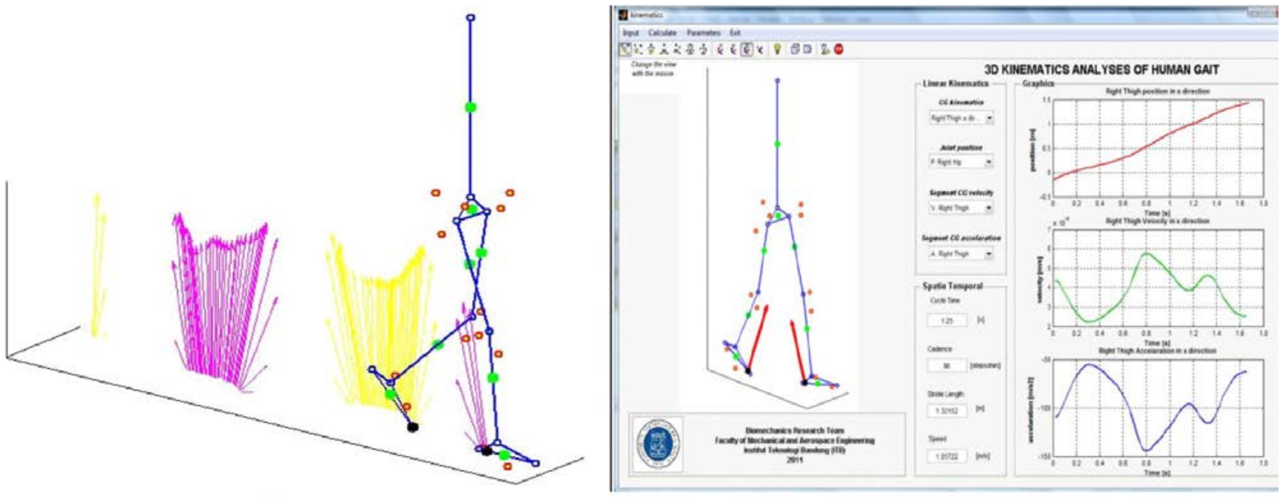
where $P\ heel_{(t)}$ is heel position, $P\ toe_{(t)}$ is toe position, and Ts is heel strike to toe off duration. Notice that *t* = 0 at heel strike and *t* = *Ts* at toe off.

After GRFs are obtained, the forces of each joint could then be calculated. The calculation is started from all distal to proximal segments. The force vector acted in proximal end of each segment is computed by knowing force vector at distal end and acceleration vector of the segment center of mass [19].

To visualize calculation results, ground forces are animated using the developed program. The results are depicted in Figure 5a. The figure also shows CoP (black dot) and CoM (green dot). In this work, a graphical user interface (GUI) is also constructed to execute program and to show results readily. The GUI is shown in Figure 5b.

Result and Discussion

The software developed in the present work is used to calculate general gait/spatio temporal parameters (cycle time, cadence, stride length and speed), linear kinematics parameters, angular kinematics, GRFs and joint forces. The obtained parameters are presented in the subsequent sub-chapters.



a) b)

Figure 5. a) Simulation of human walking b) Graphical user interface (GUI) developed in the present work

Spatio Temporal

In order to calculate spatio temporal parameters, all the primary parameters such as distance, time, number of steps, and walking phase, should be calculated first. After that, the following parameters could be obtained as follows:

1. Cycle time (s) = time (s) × 2(steps counted).
2. Cadence (steps/min) = (steps counted)× 60/time (s).
3. Stride length (m) = distance (m) × 2(steps counted).
4. Speed (m/s) = distance (m)/time (s).

The results are presented in Table 1.

Table 1. Spatio Temporal Subject

Cadence (Steps/Min)	Cycle time (s)	Stride length (m)	Speed (m/s)
96	1.25	1.3	1.03

Linear Kinematics

At present, the program could calculate linear kinematics parameter such as joint/segment position, its velocity and linear acceleration. Due to space restriction not all segment parameters are shown in this paper. Figure 6 shows right foot segment positions, velocities and accelerations in X, Y, and Z direction.

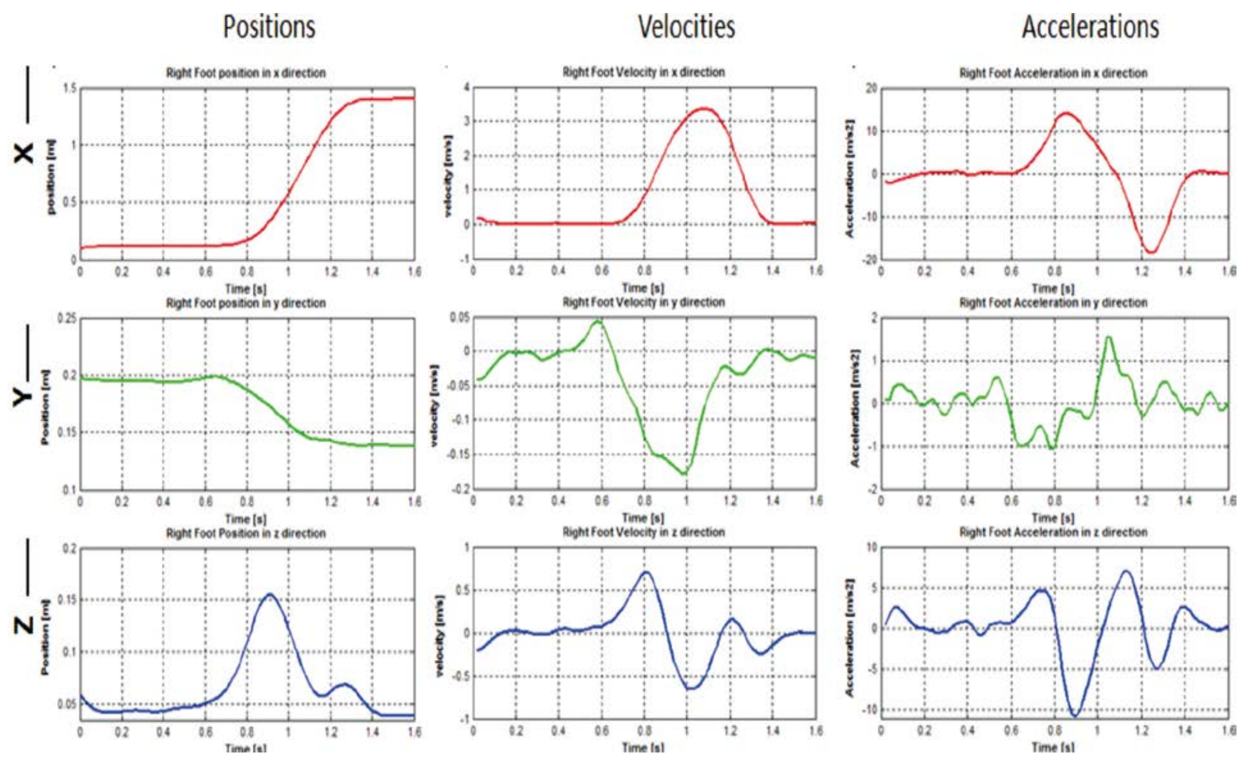


Figure 6. Positions, velocities and accelerations of right foot in X- Y- Z direction

Angular Kinematics

Angular kinematics parameters calculated by the program are joint angles, angular velocities and accelerations. Those parameters are depicted in Figure 7. It should be noted that the angular velocities-accelerations are relative to anatomical axes of each segment.

Kinetics

The developed software could successfully calculate GRFs and joint forces. The results are shown in Figure 8. The GRFs component of right and left foot during walking are presented in Figure 8a, whereas Figure 8b shows the GRFs vectors initiated from CoP.

Forces of ankle, knee and hip joint during one complete gait cycle normalized by body mass are shown in Figure 8c. It may be seen that the joint forces show similar pattern with the GRFs.

All kinematics, GRFs and joints forces parameters obtained have been compared with others published data. For spatio temporal parameters, the results agree very well with [20] for 18-49 age-group male subjects. As for linear kinematics, all calculated parameters are in a good agreement with reference [8], as well as angular kinematics parameters compared to [21], [8] and [14]. For kinetics parameters the results are showing similar pattern and comparable with [18] and [21].

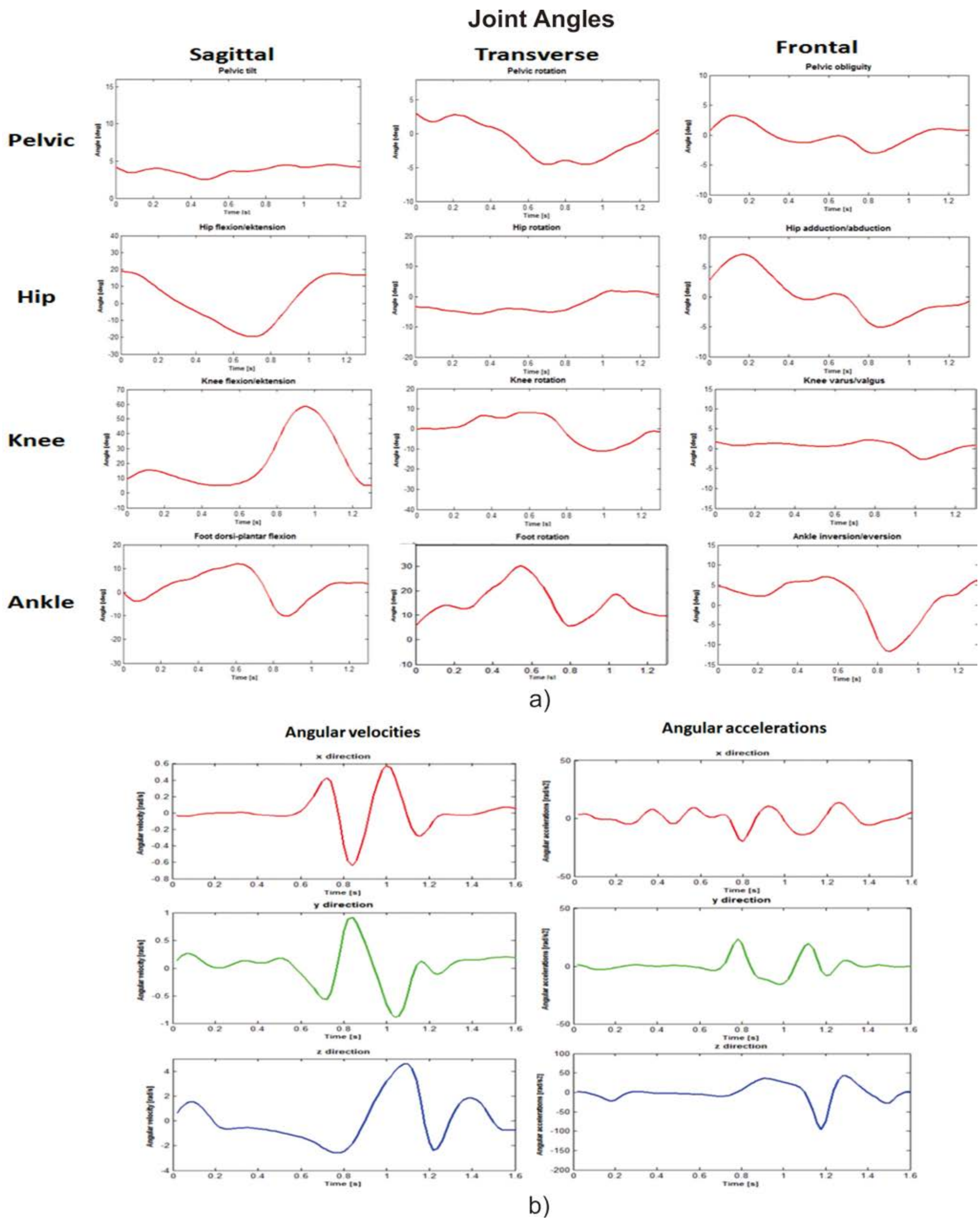


Figure 7. a) Joint angles in sagittal, transverse and frontal planes during one gait cycle
b) Angular velocities and accelerations of right calf relative to x-y-z of anatomical axes

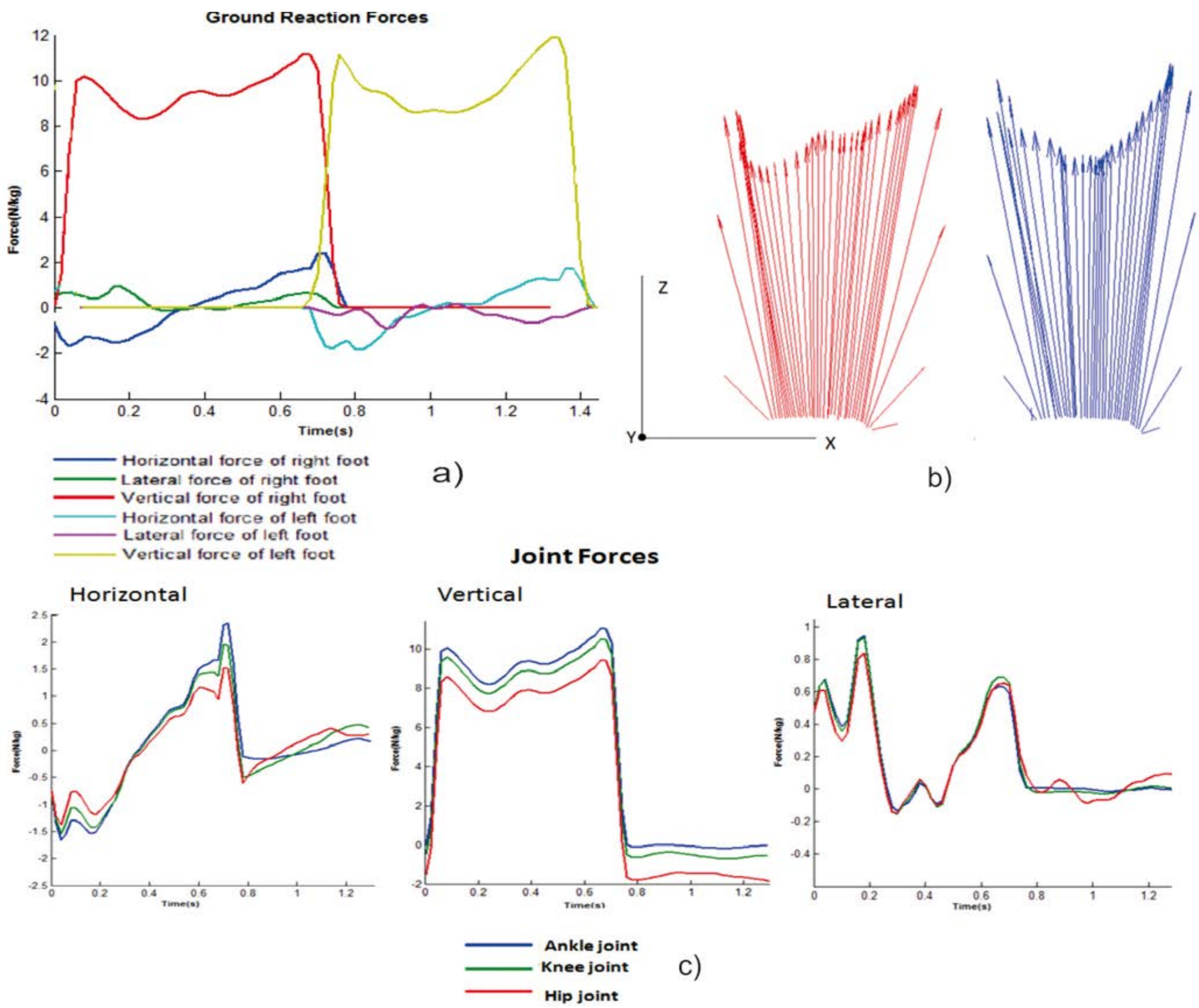


Figure 8. a) Calculated ground reaction forces of left and right foot normalized by body mass b) Ground reaction force vector of right (red arrows) and left (blue arrows) foot. Vector is initiated from CoP c) Forces of ankle, knee and hip joint during one complete gait cycle normalized by body mass

Conclusions

In this paper, a software to process and display *3D* kinematic and kinetic parameters of human walking based on position and displacement of external markers has been presented. The software could successfully calculate spatio temporal, linear and angular kinematics parameters, GRFs and joint forces of human walking.

To ensure the reliability of the developed software, the calculated parameters have been compared with those available from the literature. The comparison showed that the software is sufficiently accurate and feasible to use.

Future works include the integration of the software developed in the present work with *3D* motion capture system developed by Ferryanto [22] to establish an integrated *3D* gait analyzer system.

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