Inverse Dynamics of Change of Direction Manoeuvres in Elite Athletes

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October 2022

Abstract

Performance enhancement and injury prevention are key to the success of elite athletes. Biomechanics is a valuable tool to provide insight into the behaviour of the musculoskeletal system during the execution of sport specific tasks. The aim of this work is to assess muscle force of the human lower limbs during the execution of handball specific changes of direction (CODs) (45°, 90° and 135°). Eleven elite female handball players performed 6 COD tasks (3 with each limb). Musculoskeletal models were scaled and used to execute inverse kinematic and inverse dynamic analyses in OpenSim. Static optimization was utilised to estimate muscles' activity and force. Statistical parametric mapping was employed to compare average joint angles, joint moments, and muscle forces between the different COD angles. Sharper CODs were found to be associated with steeper deceleration and smaller hip flexion moments. The 90° CODs were found to have the smallest knee flexion and largest knee extension joint moments, together with the smallest ankle plantarflexion and largest dorsiflexion joint moments. The *rectus femoris*, the *adductors*, the *psoas* and the *tibialis anterior* muscles showed largest peak forces for the 90° CODs, while the *soleus*, the *gastrocnemius* and the *tibialis posterior* muscles showed the smallest peak forces for those CODs. All the remaining muscles showed decreasing peak force with the increase in COD angle. Statistical comparisons between dominant and non-dominant limbs showed no significant difference between them, regardless of the evaluated variables.
Keywords: Change of direction, Musculoskeletal modeling, Inverse kinematics, Inverse dynamics, Static optimisation

1. Introduction

Fully understanding a sports injury, how to prevent it and how to rehabilitate athletes to the performance level they had before the injury requires extensive knowledge of not only the risk factors and injury mechanisms, but also the athletes' execution of specific tasks as well. The description of the biomechanics by which the injury occurs is critical to identifying which intrinsic factors and which specific tasks put the athletes more at risk.[1]

Knee injuries are the most prevalent in sports.[1] The most debilitating of the knee injuries is the rupture of the ACL, which still poses the greatest challenge in sports medicine.[1] Non-contact ACL injuries have clearly been associated with twisting impacts, where the joint's internal or external rotation and hyperextension play a definitive role. Motions like landing from jumps and suddenly changing direction while running carry the most risk for this type of injury.[1] Therefore, athletes that practice sports like football, rugby or handball are particularly susceptible. Furthermore, the ratio between female and male athletes that sustain ACL injuries is about 2.6, which implies women as the most vulnerable population.[1]

Several papers have focused on the relationship between ACL injury and the biological sex of the athletes. According to Hahn et al.(2021), regardless of the type and amount of sports practiced by athletes, knee instability is very common, especially among women.[2] A review of scientific papers regarding handball injuries, by Raya-González et al. (2020), reported that male athletes most commonly injured the knees and ankles, whereas female athletes primarily injured the knees.[3] This means that not only is it more common for female athletes to sustain knee injuries than male athletes, but also that it is the most prevalent handball related injury in women. Impairments of the motion of the hip have been reported by Powers (2010) to strongly affect the kinematics and dynamics of the knee joint. He also suggested females may be more susceptible to this influence than males, potentiating ACL injury risk.[4] A review paper by Bencke et al. (2018) on gender disparities of muscle activation during the performance of sports' tasks greatly associated with ACL injury found that young female athletes present reduced activity of hamstring muscles and elevated activity of the quadriceps when compared to men.[5]

Pappas et al. (2013) reported several risk factors for ACL rupture, such as limb asymmetry, excessive quadriceps forces, excessive knee valgus and poor trunk control. This information indicates that motions that potentiate these characteristics, such as CODs, carry higher risk for ACL tear.[6] The risk factors associated with sports injuries were also investigated by Van Der Does et al. (2016) in a kinematic and dynamic study of countermovement jump landings across female and male korfball, volleyball and basketball players. They reported that landing with smaller knee flexion moments, combined with greater ground reaction forces, carries higher risk for knee injury through overuse, and that greater ankle dorsiflexion moments increases risk of acute injury of the ankle.[7] Therefore, studying how athletes perform these manoeuvres can provide key information for injury prevention and rehabilitation. The need for skill-specific training of high risk maneuvers like CODs, with the purpose of correcting mechanical behaviours of the lower limbs joints, was highlighted by Weiss and Whatman (2015) as an effective way to reduce ACL injury risk.[8] This claim is supported by Dos'Santos et al. (2019) in a review paper that states that the most effective training programs to reduce knee injury risk consisted of balance training and changing of the COD technique.[9]

The question then arises as to whether the best for injury management is also the best for performance. Dos'Santos et. al (2018) explored the influence of angle and velocity in COD manoeuvres in sports and the relation of these parameters to a performance-injury conflict. They reported that the most desirable angles and speeds for athletic success also carry the most risk for injury. Identifying the existence of a necessary trade-off between approach velocity and the ability to execute the COD manoeuvre is one of the highlights of this analysis. Furthermore, it reinforced the need for further biomechanical studies considering the impact of these two parameters in the performance of COD motions.[10] This theme was further investigated by Fox (2018). The key takeaway from this study is that prevention programs must take into account the performance of the COD task, to make sure they can be used without compromising athletic success.[11] Different types of training and their effects on COD execution were analysed by Nygaard et al. (2019), reporting that their effectiveness is strongly dependent on the type of the COD task.[12] This study reports the need to adapt training to the specific requirements of the manoeuvres, highlighting the importance of knowing the biomechanical specificities of each sports' CODs, in order to develop optimal training procedures.

Angle dependence of the biomechanical behaviour of lower limb joints during COD motions was the focus of another work developed by Dos'Santos et al. (2021), supporting the previously mentioned notion that sharper changes of direction are riskier for the knee. This work also states that the use of biomechanical analysis of these motions, to assess injury risk or to help in training, must consider the COD angle, since it affects joint movement patterns.[13] Findings by João et al. (2014)[14] support the notion of synergies between joints (the knee and ankle in this case) in the control of complex motions, such as hopping or changing direction. Therefore, to better understand the biomechanical behaviour of the lower limbs during sports related tasks, the various joints acting together must be considered.

Considering the sport specificity and the other aforementioned factors that influence COD performance, the biomechanical analysis of this manoeuvre requires the usage of reproducible tasks. By describing the differences in kinematic and dynamic behaviour of the knee between drop-jump and COD tasks in elite female handball athletes, Kristianslund and Krosshaug (2013) and, later, Later, Nedergaard et al. (2020), concluded that drop-jump motions cannot be used to infer about knee torques in CODs.[15,16] These finds support the need for specific tasks to be designed to emulate the in-game situations of interest.

The 5-0-5 agility test has been used as a reliable substitute task to evaluate a person's general ability to change direction rapidly. During this test, the subject is instructed to sprint in a straight line during 15 meters, at which point he must turn 180 degrees and sprint back another 5 meters. Although this task is highly reproducible, adjustments must be made in order to better represent sport specific COD tasks.[17] Variations of the 5-0-5 agility test have been used across studies to represent CODs, and most importantly, handball-specific CODs.[18–20]

Other factors, such as fatigue level and chronic pains, have also been reported to influence the ability of athletes to perform COD manoeuvres and they may lead to different adaptation strategies for each subject, which impact the kinematics of the COD, as stated by Franklyn-Miller et al. (2017) and Hosseini et al. (2020). This underlies the need to try to minimise the influence of such factors when designing a biomechanical study with the objective of analysing CODs in a population of athletes.

Several studies have been performed with the intent of estimating muscle participation in motor tasks, with human gait being the one of the most studied.[21–24] Arnold et al. (2005) used a musculoskeletal model with 54 muscle-tendon actuators to compute muscle forces of the lower limb during gait.[22] The same type of model and strategy was adapted by Shelburne et al. (2005) to estimate muscle, ligament and joint forces at the knee during walking,[25] and by Sritharan et al. (2012) to compare the contribution of knee-spanning and non-knee-spanning muscles on knee loading during gait.[26] One critical factor to consider when analysing kinematic and dynamic results obtained from these analysis is their great dependency on subject specific anthropometric and geometric data, as reported by Dao et al. (2009), reinforcing the importance of scaling the model according to subject specific parameters.[27] The capability of this type of actuated musculoskeletal models to analyse human gait, calculate joint angle variations, joint moments and muscle forces becomes the starting point for studying more complex motions.

To facilitate musculoskeletal based analysis, several software have been developed for research and commercial purposes. Among the available software, OpenSim, a free, open source software that allows for the development of musculoskeletal models and dynamic simulations of a multitude of motions, is one of the most used.[28] It provides the tools to design a model of a desired structure, its actuators, and perform a series of analysis, such as inverse kinematics and inverse dynamics.[29,30] Hamner et al. (2011) successfully used it to develop a muscle-actuated musculoskeletal model to evaluate muscle contribution in propulsion and support of a male subject when running.[31] Maniar et al. (2017) utilised a freely available muscle-actuated musculoskeletal model[32] and the capabilities of OpenSim[29,30] (inverse kinematics, inverse dynamics and static optimisation) to estimate the participation of different lower limb muscles during side-step cutting manoeuvres (not sport related) in healthy subjects and assess their individual role in loading the ACL. This study reported that the hamstring and gluteal muscles contribute the most to protecting the ACL, by opposing the forces it sustains.[33,34] A fully computational strategy to estimate muscle forces was employed by Schache et al. (2010) in order to understand the loads in the hamstring muscles during sprinting and their relation to the risk of injuring said muscles. The study consisted of inverse dynamic analysis of one australian rules football player. It was performed utilizing an actuated musculoskeletal model[22] to report joint angles, moments and muscle forces of the lower limbs.[35] The OpenSim[29,30] software and its biomechanical analysis tools were also used in a study by Mateus et al. (2020) aimed at describing lower limb muscles' role in joint behaviour during abrupt deceleration tasks. Abrupt deceleration can be viewed as a part of several manoeuvres elite level athletes perform, including CODs. This work analysed elite male indoor-sports players using a muscle driven analysis in OpenSim,[29,30] referred to as computed muscle control, which is a dynamic optimisation technique to estimate muscle forces.[36] The obtained results were also compared to those computed through static optimisation, showing no significant differences between the two strategies.[36] Another comparison between the computed muscle control and the static optimisation methods was done by Roelker et al. (2020). The reported results showed that both methods' accuracy to experimental data obtained for gait varied according to specific muscle and was dependent on the musculoskeletal model used. The main conclusion was that no method was objectively better than the other and therefore neither should be used in detriment of the other.[21]

Time continuous statistical analysis have been employed in biomechanical studies to assess differences between limbs, populations and interventions.[37] It has also proven useful in studying sports injury and performance.[38,39] Results from a study by Whyte et al. (2018), utilised SPM, suggest hip and knee control plays an important role in the execution of COD manoeuvres.[40] Thomas C. Dos'Santos et al. (2021) also employed SPM to asses inter-limb differences in joint kinematics and kinetics in female football players changing direction to 180○. They found no significant differences between COD execution with dominant and non-dominant limbs on their population of athletes.[41] These studies used kinematic and dynamic analysis of experimentally acquired motion data to perform the SPM.[39–41]

2. Methodology

2.1. Musculoskeletal Modelling

This work used the Hill-type model to represent muscle.[42] This model is composed of a contractile component and an elastic component connected in parallel with each other, and another elastic component connected in series. The contractile component accounts for the active force produced by the muscle-tendon unit. More specifically, it is responsible for the contraction dynamics of the muscle. The elastic element in series represents the passive force production of tendons, and is rendered as a non-linear spring, in an effort to reproduce the stress-strain behaviour correctly. The elastic element in parallel represents the elasticity of connective tissue and intrinsic components of muscle, which present some resistance to stretching when the muscle is not activated. This model of the muscle-tendon unit is shown in Figure 1.[43,44] In order to properly model the mechanical properties of different muscles and estimate their active forces, the muscle's optimal fibre length (${l^M}_0$), its peak isometric active force (${F^M}_0$), maximum shortening velocity (v_{max}) and pennation angle for when the fibres are at optimal length (optimal pennation angle (α_0)) must be known.[32,45]

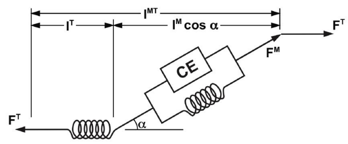

Figure 1: Hill-type computational model of muscle-tendon units in equilibrium.[32]

2.2. Biomechanical Model

The general musculoskeletal model used in this work was developed by Rajagopal et al. (2016)[32] and consists of 22 articulating rigid bodies actuated by 80 massless MTU actuators (40 in each leg) and 17 ideal torque actuators (upper body), resulting in 35 DOFs. The reference frames for joint origins and DOFs of the most relevant joints for this work are presented in Figure 2. The generic MTU actuators' geometry and placement are also shown in Figure 2.

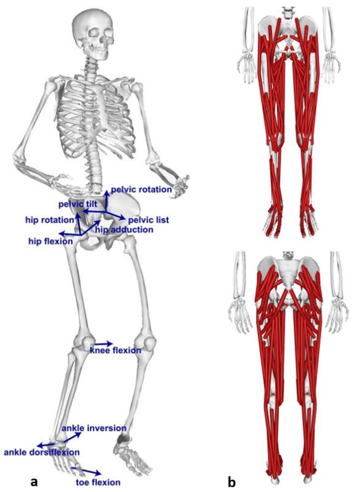

Figure 2: Representation of the rigid bodies and DOFs in the model, labeled for the right limb (a) and the MTU actuators of the model's lower limbs (b).[32] Image obtained with OpenSim.

2.3. Inverse Kinematics

In this case, the IK tool in OpenSim was used to calculate joint angles and translations during the execution of the CODs. It works by forcing the model, respecting its joint's DOFs, to assume the position that better corresponds to the experimental marker coordinates obtained from the motion capture system, at each time frame. This is achieved by solving of a weighted least squares problem (see Equation 1) The objective is to find the vector of generalised coordinates,$\boldsymbol{q}$, that minimises the distance between the experimental marker's position and the position of the marker on the model (marker error), as well as the difference between the experimental coordinate values (i.e. joint angles calculated outside OpenSim) and the coordinate values estimated by the IK (coordinate errors).[29] Since no experimental coordinate values were inputted to OpenSim, the cost function used in IK was as follows:

$$f = \sum_{i=0}^{markers} \omega_i \left(\boldsymbol{x}_i^{exp} - \boldsymbol{x}(\boldsymbol{q})_i^{model} \right)^2 \tag{1}$$

where $\boldsymbol{q}$ is the vector of generalised coordinates, ω_i is the weight factor and $\boldsymbol{x}_i^{subject}$ and $\boldsymbol{x}_i^{model}$ are the three-dimensional positions of the i^{th} marker for the subject and model, respectively.[29] The weight factors indicate how strongly the error term of a specific marker must be minimised. For this work, weight factors were the same as those used by Rajagopal et al. (2016) in their running simulations.[32]

2.4. Inverse Dynamics

ID is the process by which it is possible to calculate the internal forces that produce a certain movement, by systematically solving the EoM of a system. In order to do that, anthropometric measurements, mass, inertia and location of the centre of mass must be known for every segment of the system. The external forces resulting from the movement of the system and the kinematic variables are also necessary for this process. The EoM that describe a motion can be written as follows:

$$\boldsymbol{M}(\boldsymbol{q})\ddot{\boldsymbol{q}} + \boldsymbol{C}(\boldsymbol{q}, \dot{\boldsymbol{q}}) + \boldsymbol{G}(\boldsymbol{q}) = \tau \tag{2}$$

where $\boldsymbol{q}, \dot{\boldsymbol{q}}, \ddot{\boldsymbol{q}} \in \mathbb{R}^N$ are the vectors of generalised positions, velocities and accelerations, respectively, $\boldsymbol{M}(\boldsymbol{q}) \in \mathbb{R}^{NxN}$ is the system mass matrix, $\boldsymbol{C}(\boldsymbol{q}, \dot{\boldsymbol{q}}) \in \mathbb{R}^N$ is the vector of Coriolis and centrifugal forces, $\boldsymbol{G}(\boldsymbol{q}) \in \mathbb{R}^N$ is the vector of gravitational forces and $\tau \in \mathbb{R}^N$ is the vector of generalised forces, with N representing the total number of DOFs in the system.[29]

2.5. Muscle Force Estimation

The SO tool is an extension of the ID tool used to estimate individual muscle forces acting on the system to produce its motion. It does so by decomposing the net joint forces and torques into individual muscle forces, while taking into account force-length-velocity properties of the MTU actuators (Equation 3).

$$\sum_{m=1}^{n} \left[a_m f(F_m^0, l_m, v_m) \right] r_{m,j} = \tau_j \tag{3}$$

where n is the number of MTUs in the model, a_m is the activation of the m^{th} muscle at a given time step, F_m^0, l_m and v_m are its maximum isometric force, length and shortening velocity, respectively, $f(F_m^0, l_m, v_m)$ is its force-length-velocity surface, $r_{m,j}$ is its moment arm about the j^{th} joint axis and τ_j is the generalised force acting about that joint axis.

This generates an under-determined system of equations, as there are more unknowns than the number of DOFs in the model, which results in an infinite number of solutions. This optimisation strategy surpasses this problem by finding the solution that minimises a specific objective function. In this work, the SO was instructed to solve the EoM for the generalised forces while minimising the squared sum of their activations (Equation 4).

$$J = \sum_{m=1}^{n} (a_m)^2 \tag{4}$$

This process is repeated independently for each timestep of the analysed motion.

2.6. Statistical Analysis

After obtaining the outcomes from IK, ID and SO, for every trial of the 11 subjects, the output files of interest were post-treated in Matlab. All results were low-pass filtered, using a Butterworth filter with a cut-off frequency defined through Residual Analysis. They were normalised in the time-domain and ploted in terms of percentage of COD completion, so that 0% corresponds to the first instant of contact with the force plate and 100% to the last. Joint moments and muscle forces were also normalised by the mass of the subjects.

SPM was used to assess inter-limb differences between the total average results on the 3 COD angles.[39,41] It was also implemented to statistically compare the total average results across the different COD angles, in order to investigate angle dependence of the variables of interest.[37] In this case, 2-tailed paired t-tests were used to compare pairs of results, and the null-hypothesis corresponds to no difference between these results. The defined statistical significance of $\alpha = 0.05$ signifies that for every instant the SPM produces a t value that surpasses the critical threshold, t^*, the null-hypothesis is rejected with a confidence level of 95%. SPM produces p-values across the full time-series, indicating the probability of the supra-threshold cluster being as large (in the time domain) as the observed cluster in the statistical test.[46,47] SPM analysis was performed using the open-source Matlab package *spm1d* (spm1d.org, ©T. Pataky)

2.7. Experimental Procedure

Eleven healthy female handball players (age: 22 ± 4 years, height: 1.69 ± 0.08 m, weight: 67 ± 12 kg) participated in this study. The acquisition of three-dimensional marker data was carried out in the Laboratory of Biomechanics of Lisbon (Instituto Superior Técnico) using a motion capture system (MOCAP) composed of 14 infrared ProReflex 1000 cameras (Qualisys ©, Göteborg, Sweden). GRF were also acquired, by means of 3 force plates (AMTI, OR 6-7-1000 508x464mm, Watertown, MA). The acquisition frequency of the MOCAP system and of the force plates was set to 100 Hz for both. 71 reflective markers were placed on the subjects (47 on anatomical prominences and 24 clusters).[33]

The task design consisted of three CODs, at 45°, 90° and 135° angles. Each COD was performed using both the left and right legs. Subjects were running in place at the start of a 1m line placed on the floor. Upon a signal, they took a small support step and then a second step along that line in order to place the stance foot on the centre of the force plate. Then they changed directions to sprint along another line marked on the floor, at the desired angle. Figure 3 illustrates the task design. 5 successful trials were acquired for each COD, making a total of 30 dynamic trials per subject.

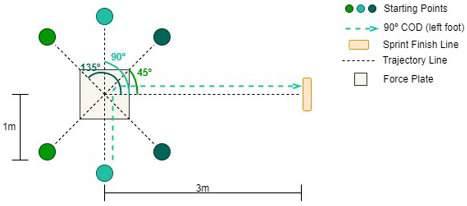

Figure 3: Schematic representation of the laboratory setup and the execution of the 90° COD with the left foot.

All experimental data was filtered using low-pass Butterworth filters with cut-off frequencies calculated through residual analysis, as proposed by Winter (2009).[48]

3. Results & Discussion

The most relevant results from IK, ID and SO are summarised in Tables 1, 2 and 3, respectively. The average maximum and minimum velocities, maximum and minimum joint angles, maximum and minimum joint moments, maximum and minimum estimated muscle forces and their respective timings are presented for the 3 COD angles, executed with the dominant and non-dominant limb.

The curves regarding hip flexion, adduction, rotation, knee flexion and ankle flexion angles for a 45° COD obtained in this work are similar in profile and magnitude to those presented by Maniar.[33] Knee flexion and ankle dorsiflexion moments reported by Maniar[33] for the 45° CODs are also similar in shape to those obtained in this work. Regarding joint motion in the sagittal plane, hip and knee flexion angles obtained in this work are in agreement with those reported by Dos'Santos et al.[13] and Schreurs et al.[18] In both cases, flexion at IC decreased with the increase in cutting angle. Ankle dorsiflexion followed the same pattern, showing greater angles at IC for the 45° COD. The statistical analysis in this work did not identify significant differences in peak knee flexion and ankle dorsiflexion for the different COD angles. Rather, this analysis highlights IC as the COD phase where cutting angle influences the sagittal plane kinematics the most.

As another measure of the impact the cutting angle has on COD execution, this work found that sharper CODs are performed at lower velocities. Not only that, initial and terminal velocities also decrease with the increase in cutting angle. These findings are in accordance with those by Dos'Santos et al.,[10,13] which reported a trade-off between velocity and COD angle. Moreover, as COD angle increases, the WA and Acceleration phases correspond to steeper deceleration and acceleration, respectively. Mateus et al.[36] found the *Soleus* muscle to be one of the main contributors to the deceleration during tasks like CODs. Maniar et al.[34] also reported the *Gastrocnemius* to play a significant role in deceleration when changing direction. This information indicates that these muscles are expected to increase their produced force with the increasing COD angles.

Since the *Soleus* and *Gastrocnemius* muscles are both plantarflexor muscles, their behaviour reflects on the ankle joint moment. This work's statistical analysis of the ankle motion revealed plantarflexor moments to be smaller for the first 40% of the 90°COD than for the 135° COD, which is to say, during the phase where deceleration occurs. These results also found a significant increase in force production by these two muscles towards Toe-off of the 135° COD. As no deceleration is present at this time, this may indicate these muscles have another function other than braking, which is also accentuated by the increase in COD angle.

Table 1: Inverse Kinematics Results: maximum and minimum average velocities; peak hip flexion(+) and extension(-); peak hip adduction(+) and abduction(-); peak hip internal(+) and external(-) rotation; peak knee flexion(+) and extension(-); peak ankle dorsiflexion(+) and plantarflexion(-) average joint angles(°). Respective timings presented in % of COD execution. Results presented as average ± standard deviation, for all COD angles, for the dominant (top) and non-dominant (bottom) limbs.

Inverse Kinematics						
Dominant Limb						
	45°		**90°**		**135°**	
	Vel. (m/s)	% COD	Vel. (m/s)	% COD	Vel. (m/s)	% COD
Velocitiy (Max/Min)	3.4 ± 0.5 / 2.6 ± 0.4	100% / 0%	2.5 ± 0.4 / 1.6 ± 0.3	100% / 45%	2.0 ± 0.4 / 0.9 ± 0.4	100% / 45%
	Angle (°)	% COD	Angle (°)	% COD	Angle (°)	% COD
Hip Flexion-Extension (Max/Min)	50.0 ± 6.7 / −16.7 ± 7.8	0% / 100%	38.9 ± 8.3 / −14.4 ± 8.3	0% / 100%	28.9 ± 6.7 / −13.3 ± 8.9	0% / 100%
Hip Addcution-Abduction (Max/Min)	−6.0 ± 6.0 / −15.5 ± 3.5	36% / 88%	−15.8 ± 5.0 / −26.0 ± 6.5	0% / 73%	−22.0 ± 7.0 / −27.0 ± 8.0	100% / 67%
Hip Internal-External Rotation (Max/Min)	4.4 ± 10.0 / −12.5 ± 7.5	33% / 100%	−4.4 ± 8.8 / −23.1 ± 3.8	9% / 93%	2.5 ± 5.0 / −26.3 ± 3.8	9% / 93%
Knee Flexion-Extension (Max/Min)	52.1 ± 4.3 / 17.1 ± 3.6	37% / 100%	−53.6 ± 5.7 / −20.7 ± 5.7	45% / 92%	53.6 ± 7.9 / −19.3 ± 2.9	47% / 92%
Ankle Dorsiflexion-Plantarflexion (Max/Min)	22.1 ± 3.6 / −9.3 ± 4.3	51% / 100%	−19.6 ± 4.3 / −20.7 ± 7.9	57% / 100%	16.4 ± 5.7 / −24.3 ± 3.6	47% / 100%
Non-dominant Limb						
	45°		**90°**		**135°**	
	Vel. (m/s)	% COD	Vel. (m/s)	% COD	Vel. (m/s)	% COD
Velocitiy (Max/Min)	3.4 ± 0.7 / 2.6 ± 0.6	100% / 0%	2.5 ± 0.3 / 1.7 ± 0.2	100% / 45%	2.0 ± 0.3 / 0.8 ± 0.3	100% / 45%
	Angle (°)	% COD	Angle (°)	% COD	Angle (°)	% COD
Hip Flexion-Extension (Max/Min)	50.0 ± 10.0 / −15.6 ± 7.2	0% / 100%	36.7 ± 5.6 / −15.6 ± 5.6	0% / 100%	30.0 ± 7.8 / −10.0 ± 8.3	0% / 100%
Hip Addcution-Abduction (Max/Min)	−4.5 ± 5.0 / −12.5 ± 5.0	0% / 85%	−15.5 ± 6.0 / −21.0 ± 7.0	100% / 63%	−20.0 ± 7.0 / −25.5 ± 4.0	100% / 12%
Hip Internal-External Rotation (Max/Min)	3.1 ± 8.1 / −14.4 ± 10.6	24% / 100%	6.3 ± 8.1 / −22.5 ± 10.0	9% / 100%	5.6 ± 8.8 / −24.4 ± 8.1	15% / 95%
Knee Flexion-Extension (Max/Min)	54.3 ± 4.3 / −17.1 ± 3.6	41% / 100%	55.0 ± 5.0 / −18.6 ± 2.9	45% / 93%	58.6 ± 7.9 / −21.4 ± 5.7	44% / 91%
Ankle Dorsiflexion-Plantarflexion (Max/Min)	22.1 ± 3.6 / −19.3 ± 3.6	52% / 100%	22.1 ± 3.6 / −21.4 ± 4.3	49% / 100%	20.0 ± 5.0 / −23.6 ± 4.3	47% / 100%

Table 2: Inverse Dynamics Results: peak hip flexion(+) and extension(-); peak hip adduction(+) and abduction(-); peak hip internal(+) and external(-) rotation; peak knee flexion(+) and extension(-); peak ankle dorsiflexion(+) and plantarflexion(-) average joint moments (Nm/Kg). Respective timings presented in % of COD execution. Results presented as average ± standard deviation, for all COD angles, for the dominant (top) and non-dominant (bottom) limbs.

Inverse Dynamics						
Dominant Limb						
	45°		**90°**		**135°**	
	Moment (Nm/Kg)	% COD	Moment (Nm/Kg)	% COD	Moment (Nm/Kg)	% COD
Hip Flexion-Extension (Max/Min)	1.2 ± 0.4 / −1.7 ± 0.4	100% / 0%	1.1 ± 0.3 / −1.1 ± 0.4	100% / 0%	0.9 ± 0.3 / −1.9 ± 0.3	100% / 0%
Hip Addcution-Abduction (Max/Min)	1.4 ± 0.9 / −0.0 ± 0.2	56% / 0%	1.8 ± 0.8 / 0.2 ± 0.2	61% / 0%	1.5 ± 1.1 / 0.6 ± 0.3	63% / 100%
Hip Internal-External Rotation (Max/Min)	0.0 ± 0.1 / −0.5 ± 0.4	0% / 39%	−0.1 ± 0.1 / −0.7 ± 0.4	100% / 50%	−0.1 ± 0.1 / −0.5 ± 0.5	100% / 43%
Knee Flexion-Extension (Max/Min)	0.7 ± 0.3 / −2.0 ± 1.3	0% / 44%	0.4 ± 0.1 / −2.9 ± 0.8	0% / 43%	0.7 ± 0.2 / −1.8 ± 1.0	0% / 36%
Ankle Dorsiflexion-Plantarflexion (Max/Min)	−0.1 ± 0.1 / −2.6 ± 0.8	0% / 64%	0.0 ± 0.1 / −1.9 ± 0.4	0% / 67%	−0.3 ± 0.1 / −2.4 ± 0.4	0% / 67%
Non-dominant Limb						
	45°		**90°**		**135°**	
	Moment (Nm/Kg)	% COD	Moment (Nm/Kg)	% COD	Moment (Nm/Kg)	% COD
Hip Flexion-Extension (Max/Min)	1.3 ± 0.4 / −1.8 ± 0.4	100% / 0%	1.2 ± 0.3 / −1.3 ± 0.4	100% / 0%	1.0 ± 0.2 / −1.0 ± 0.2	100% / 0%
Hip Addcution-Abduction (Max/Min)	0.8 ± 1.0 / −0.1 ± 0.3	56% / 0%	1.3 ± 1.0 / 0.2 ± 0.3	34% / 0%	1.0 ± 0.3 / 0.6 ± 0.2	100% / 0%
Hip Internal-External Rotation (Max/Min)	0.0 ± 0.2 / −0.3 ± 0.3	0% / 41%	−0.1 ± 0.1 / −0.6 ± 0.4	100% / 43%	0.1 ± 0.1 / −0.4 ± 0.3	100% / 40%
Knee Flexion-Extension (Max/Min)	0.7 ± 0.2 / −2.4 ± 1.2	0% / 46%	0.7 ± 0.2 / −3.6 ± 1.2	0% / 43%	0.6 ± 0.1 / −2.4 ± 0.9	0% / 34%
Ankle Dorsiflexion-Plantarflexion (Max/Min)	0.0 ± 0.2 / −2.4 ± 0.8	0% / 64%	0.0 ± 0.1 / −1.8 ± 0.6	0% / 66%	−0.3 ± 0.1 / −2.0 ± 0.6	0% / 66%

Table 3: Static Optimization Results: maximum and minimum *Gluteus maximus*, *Gluteus medius*, *Gluteus minimus*, *Psoas*, *Piriformis*, *Adductor brevis*, *Adductor longus*, *Semimembranosus*, *Semitendinosus*, *Tibialis anterior*, *Tibialis posterior*, *Gastrocnemius* and *Soleus* average force (N/Kg). Respective timings presented in % of COD execution. Results presented as average ± standard deviation, for all COD angles, for the dominant (top) and non-dominant (bottom) limbs.

Static Optimization						
Dominant Limb						
	45°		**90°**		**135°**	
	Force (N/Kg)	% COD	Force (N/Kg)	% COD	Force (N/Kg)	% COD
Gluteus maximus (Max/Min)	3.0 ± 1.4 / 0.0 ± 0.0	20% / 100%	2.1 ± 0.4 / 0.0 ± 0.0	0% / 100%	1.3 ± 0.4 / 0.0 ± 0.0	13% / 100%
Gluteus medius (Max/Min)	1.5 ± 0.6 / 0.0 ± 0.0	0% / 100%	0.8 ± 0.0 / 0.0 ± 0.0	0% / 100%	0.8 ± 0.0 / 0.0 ± 0.0	30% / 100%
Gluteus minimus (Max/Min)	0.3 ± 0.0 / 0.0 ± 0.0	0% / 100%	0.0 ± 0.0 / 0.0 ± 0.0	0% / 100%	0.0 ± 0.0 / 0.0 ± 0.0	0% / 100%
Psoas (Max/Min)	12.9 ± 2.1 / 0.9 ± 2.7	100% / 0%	15.0 ± 1.8 / 1.8 ± 0.9	100% / 0%	13.8 ± 2.5 / 2.1 ± 1.3	100% / 0%
Piriformis (Max/Min)	2.7 ± 1.0 / 0.0 ± 0.0	20% / 100%	1.7 ± 0.4 / 0.0 ± 0.0	0% / 87%	0.8 ± 0.8 / 0.0 ± 0.0	0% / 79%
Adductor brevis (Max/Min)	4.1 ± 2.3 / 0.5 ± 0.0	65% / 0%	5.0 ± 1.7 / 0.8 ± 0.4	64% / 0%	3.3 ± 2.1 / 1.3 ± 0.4	74% / 0%
Adductor longus (Max/Min)	7.7 ± 2.7 / 0.5 ± 0.0	84% / 0%	7.9 ± 2.1 / 1.3 ± 0.8	78% / 0%	6.3 ± 1.7 / 1.7 ± 0.4	87% / 0%
Rectus femoris (Max/Min)	15.9 ± 8.3 / 1.7 ± 0.8	54% / 0%	26.2 ± 4.6 / 2.3 ± 0.8	48% / 0%	16.7 ± 10.0 / 0.8 ± 0.8	37% / 0%
Semimembranosus (Max/Min)	10.0 ± 2.5 / 1.7 ± 0.8	0% / 100%	8.3 ± 2.5 / 0.8 ± 0.0	0% / 100%	7.5 ± 2.5 / 1.7 ± 0.8	0% / 100%
Semitendinosus (Max/Min)	2.3 ± 0.0 / 0.0 ± 0.0	19% / 100%	1.7 ± 0.8 / 0.0 ± 0.0	64% / 100%	1.7 ± 0.8 / 0.8 ± 0.0	60% / 100%
Tibialis anterior (Max/Min)	2.9 ± 0.0 / 0.0 ± 0.0	100% / 57%	4.3 ± 0.0 / 0.0 ± 0.0	100% / 71%	2.9 ± 0.0 / 0.0 ± 0.0	100% / 64%
Tibialis posterior (Max/Min)	10.0 ± 2.9 / 0.0 ± 0.0	73% / 0%	7.1 ± 4.3 / 0.0 ± 0.0	74% / 0%	8.6 ± 2.9 / 0.0 ± 0.0	78% / 0%
Gastrocnemius (Max/Min)	15.7 ± 8.6 / 2.9 ± 1.4	61% / 0%	14.3 ± 4.3 / 1.4 ± 0.0	74% / 0%	17.1 ± 4.3 / 2.9 ± 1.4	75% / 0%
Soleus (Max/Min)	37.1 ± 14.3 / 0.0 ± 0.0	44% / 100%	32.9 ± 8.6 / 1.4 ± 0.0	53% / 100%	38.6 ± 12.9 / 2.9 ± 0.0	37% / 100%
Non-dominant Limb						
	45°		**90°**		**135°**	
	Force (N/Kg)	% COD	Force (N/Kg)	% COD	Force (N/Kg)	% COD
Gluteus maximus (Max/Min)	3.1 ± 1.2 / 0.3 ± 0.1	20% / 100%	2.1 ± 1.0 / 0.0 ± 0.0	0% / 100%	1.6 ± 1.5 / 0.0 ± 0.0	50% / 100%
Gluteus medius (Max/Min)	1.6 ± 0.9 / 0.3 ± 0.1	0% / 100%	0.7 ± 0.3 / 0.0 ± 0.0	0% / 100%	0.7 ± 0.8 / 0.0 ± 0.0	39% / 100%
Gluteus minimus (Max/Min)	0.4 ± 0.2 / 0.1 ± 0.0	0% / 100%	0.2 ± 0.0 / 0.0 ± 0.0	0% / 100%	0.1 ± 0.1 / 0.0 ± 0.0	41% / 100%
Psoas (Max/Min)	14.2 ± 3.0 / 0.7 ± 0.4	100% / 0%	14.7 ± 2.0 / 1.4 ± 1.0	100% / 0%	13.4 ± 1.9 / 3.9 ± 1.4	100% / 0%
Piriformis (Max/Min)	2.6 ± 0.8 / 0.4 ± 0.2	0% / 100%	1.7 ± 1.1 / 0.0 ± 0.0	35% / 96%	1.6 ± 1.5 / 0.2 ± 0.0	37% / 90%
Adductor brevis (Max/Min)	3.3 ± 1.8 / 0.3 ± 0.1	65% / 0%	4.9 ± 2.2 / 0.8 ± 0.1	61% / 0%	3.0 ± 1.7 / 1.5 ± 0.2	68% / 0%
Adductor longus (Max/Min)	6.6 ± 2.7 / 0.4 ± 0.1	65% / 0%	7.6 ± 2.7 / 1.0 ± 0.0	77% / 0%	5.4 ± 2.1 / 2.6 ± 0.1	87% / 0%
Rectus femoris (Max/Min)	14.2 ± 7.1 / 1.3 ± 0.9	48% / 0%	26.1 ± 8.4 / 2.4 ± 0.9	43% / 0%	19.9 ± 8.6 / 1.8 ± 0.4	33% / 0%
Semimembranosus (Max/Min)	10.1 ± 2.9 / 1.5 ± 1.0	0% / 100%	9.2 ± 2.2 / 1.1 ± 0.7	0% / 100%	6.1 ± 2.2 / 1.4 ± 0.6	0% / 100%
Semitendinosus (Max/Min)	1.5 ± 0.6 / 0.4 ± 0.1	20% / 100%	1.1 ± 1.2 / 0.0 ± 0.0	42% / 0%	1.1 ± 0.5 / 0.5 ± 0.2	17% / 100%
Tibialis anterior (Max/Min)	2.9 ± 0.0 / 0.6 ± 0.1	100% / 62%	3.5 ± 0.1 / 0.6 ± 0.0	100% / 73%	3.3 ± 0.8 / 0.4 ± 0.5	100% / 67%
Tibialis posterior (Max/Min)	9.4 ± 2.7 / 0.0 ± 0.0	73% / 0%	8.8 ± 3.4 / 0.0 ± 0.0	69% / 0%	7.2 ± 3.1 / 0.0 ± 0.0	81% / 0%
Gastrocnemius (Max/Min)	16.0 ± 5.6 / 2.2 ± 0.9	56% / 0%	12.0 ± 6.1 / 1.9 ± 0.8	67% / 0%	16.8 ± 7.4 / 3.3 ± 0.6	72% / 0%
Soleus (Max/Min)	35.1 ± 11.2 / 0.0 ± 0.0	44% / 100%	30.7 ± 9.3 / 0.0 ± 0.0	59% / 100%	40.0 ± 0.9 / 0.0 ± 0.0	40% / 100%

Regarding the knee joint, statistical comparison of the joint moments across the COD angles suggested the 135° COD presented smaller knee extensor moments at the end of Acceleration phase when compared to the 90° COD. This coincides with the observation that the *Gatrocnemius* has an increase in force around the Toe-off of the 135° COD, as it is also a knee flexor muscle, which force opposes the extensor moment. Regarding the other knee flexor muscles, *Semimembranosus* and *Semitendinosus*, the increase in COD angle mostly affects their participation during IC and Toe-off, which show a decrease in their generated force. The *Rectus femoris*, as a knee extensor muscle, appears to be responsible for the peak in knee extensor moment verified for the 90° COD, as it showed significantly higher force for the 90° COD when compared to the 135° COD during the beginning of the Acceleration phase.

These 3 thigh muscles also participate in hip flexion and extension, where the *Rectus femoris* has an opposing role to the *Semimembranosus* and *Semitendinosus* muscles. The *Psoas* and *Gluteus maximus* muscles are also involved in the control of this hip joint motion. This work found that hip flexion angles at IC decrease as COD angle increases. Similar reports were also made by Dos'Santos et al.[10] To the larger IC flexion angles verified for the 45° COD correspond larger extensor moments observed during IC of that COD, when compared with the other CODs. This alludes to an increased contribution of the hip extensor muscles during the initial phase of the 45° COD, verified by the statistical analysis.

The gluteal muscles showed a decrease their participation, particularly during the IC phase, with the increase in COD angle. These muscles were found by Maniar et al.[34] to be the largest contributors to acceleration towards the desired direction. However, the difference in the Acceleration phase of the movements showed by the velocity profiles of the various COD angles is not reflected in a significant change in the gluteal muscle's force production. Moreover, the magnitude of the gluteal forces showed during the Acceleration phases of the movements are smaller than those of other muscles, in particular, the *Psoas* and the *Adductor* muscles. This suggests these hip muscles may play a more important role in accelerating the movement towards the desired new direction. As a hip flexor, the *Psoas* contributes to approximate the trunk and thigh segments, potentiating acceleration. The 90° COD showed higher levels of force for this muscle than the other CODs, with a particularly significant increase when compared to the 45° COD. The *Adductor brevis* and *Adductor longus* muscles' function is to move the thigh (and leg) away from the mid-line of the body, in the frontal plane. As COD angle increases, so does the hip adduction moment at IC, as a result of the increased force generated by these two muscles, combined with the decrease in force produced by the hip abductor muscles (i.e. gluteal muscles). The *Adductor* muscles also show more contribution towards the Acceleration and Toe-off phases for the 90° COD than for the others. However, this is not translated in the hip adduction moment, which found no significant differences between COD angles during those phases. That is probably due the action of the abductor muscles of the hip, opposing that motion.

Since hip joint rotation is controlled by muscles which primary function is the generation of force involved in other motions, hip behaviour in this DOF becomes whatever results from the control of those other motions. The possibility of executing the CODs with an externally or internally rotated hip, as shown by the results in this work, implies that hip rotation is not critical to COD execution.

This work, similarly to Dos'Santos et al.,[41] who evaluated inter-limb variance during 180° COD tasks, found no significant differences between COD execution with the subjects' dominant or non-dominant limb. This may be an indicator that elite athletes sports involving COD manoeuvres may be less susceptible to between-limb asymmetries due to their training.

Little hip and knee flexion as well as hip internal rotation have been found to be associated with larger knee valgus, and to contribute greatly to an increase in acACL loading.[?,10,13,15] The present study found the 135° CODs to have the most dangerous combination of these factors.

4. Conclusions, Limitations & Future Work

The objective of this work was to evaluate and compare hip, knee and ankle joint angles and moments and lower limb muscle forces estimated from experimental data of female handball athletes executing COD manoeuvres with different angles. The methodology applied allowed the computation of joint kinematics and kinetics, which were shown to be consistent with the literature. Additionally, muscle activity was also computed.

The main findings pertain to the dependence of the biomechanics of the COD task on the cutting angle. Sharper CODs are associated with lower velocity profiles, steeper deceleration and acceleration and more extended hips. The COD angles differ from each other by 45 degrees. However, the differences in kinematic and dynamic behaviour of the lower limbs between the 90° and 135° CODs are generally smaller than the difference between the 45° and 90° CODs. This implies lower limb biomechanics do not have a linear response to the increase in COD angle. When it comes to knee and ankle, the change in these joint's moments is not proportional to the variation in COD angle, nor to change in kinematics. Even though the kinematics of the CODs are not as impacted when the cutting angle goes from 90° to 135°, the internal forces acting on the knee and ankle change the most with that angle increment. The 90° COD was found to be the task where the *Soleus* and *Gastrocnemius* muscles contributed less to the IC and WA phases of the manoeuvres. This is of particular importance, since these muscles have been proposed by literature to be responsible for the abrupt deceleration during the first half of the stance phase of COD tasks. The comparison between biomechanical profiles of the COD tasks executed with the subjects' dominant and non-dominant limbs revealed no significant asymmetries in the evaluated population.

This study carries a set of limitations, starting by having been performed in a laboratory setting. Even though this type of experimental acquisition allows for more accurate data to be extracted, the simplification of the task to fit within the laboratory's requirements may distance it from what would be an actual in-game situation. Furthermore, intensity and fatigue levels in the laboratory environment can be smaller than those experienced in-game, thus not accurately reflecting the events trying to be reproduced. As such, an important improvement on this work would be to investigate the possibility to measure biomechanical data in-game. This would allow for a more accurate representation of the COD tasks to be studied, and would even enable the study of other in-game situations that are not reproducible in laboratory settings.

The musculoskeletal model used also has its limitations mostly regarding degrees of freedom that could not be considered in order to successfully reproduce the desired motion. The knee and ankle joints were only considered to have movement on the sagittal plane, which is not true for natural human motion. Furthermore, the subtalar and metatarsal-phalangeal joints were considered to have no movement at all. These foot motions could play an important role in the execution of the CODs. Moreover, the muscle parameters utilized were obtained from works with different subjects, body types and strategies, and may not truly represent the subjects analysed here. Another limitation comes from the static optimization process, which considers tendons to be inextensible and therefore does not take into account muscle's parallel elastic element's contribution.

Another important limitation of this study was the small population it evaluated. Larger populations allow for more robust statistical analysis results. A great improvement this work would be to increase the population size.

It would also be of particular interest to study smaller increments in COD angle, to better understand the knee and ankle joints response. This would also account for a wider range of game-like situations.

The outcomes of this study are valuable for the development of training programs to further improve these athletes' performance while decreasing their injury risk.[13,33] Factors like increasing braking and propulsion capacity, through muscle training, as well as joint positions through technique training can be evaluated with similar studies.[15] Other than the possibility of analysing a particular athlete in detail, which is extremely valuable for elite athletes, this methodology can be applied for larger populations to identify the most common motion patterns, and introduce that knowledge in training at the younger formative levels.

5. Acknowledgement

This document was written and made publically available as an institutional academic requirement and as a part of the evaluation of the MSc thesis in Biomedical Engineering of the author at Instituto Superior Tecnico. The work described herein was performed at the Biomechanics Group of Instituto Superior Técnico (Lisbon, Portugal), during the period February 2020-October 2022, under the supervision of Prof.Dr. Carlos Miguel Fernandes Quental.

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