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Glenohumeral contact force during flat and topspin tennis forehand drives

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Abstract

The primary role of the shoulder joint in tennis forehand drive is at the expense of the loadings undergone by this joint. Nevertheless, few studies investigated glenohumeral (GH) contact forces during forehand drives. The aim of this study was to investigate GH compressive and shearing forces during the flat and topspin forehand drives in advanced tennis players. 3D kinematics of flat and topspin forehand drives of 11 advanced tennis players were recorded. The Delft Shoulder and Elbow musculoskeletal model was implemented to assess the magnitude and orientation of GH contact forces during the forehand drives. The results showed no differences in magnitude and orientation of GH contact forces between the flat and topspin forehand drives. The estimated maximal GH contact force during the forward swing phase was 3573 ± 1383 N, which was on average 1.25 times greater than during the follow-through phase, and 5.8 times greater than during the backswing phase. Regardless the phase of the forehand drive, GH contact forces pointed toward the anterior-superior part of the glenoid therefore standing for shearing forces. Knowledge of GH contact forces during real sport tasks performed at high velocity may improve the understanding of various sport-specific adaptations and causative factors for shoulder problems.

29

30 **Keywords:** Glenohumeral contact force; shoulder; musculoskeletal model; racket sports;
31 anterior-superior glenoid shearing force

32

33 **Introduction**

34 Playing tennis applies repeated high loads onto the upper limb joints, especially onto the
35 shoulder complex (Elliott, Fleisig, Nicholls, & Escamilia, 2003). If the serve places the tennis
36 player at the highest risk for impingements at the shoulder (Charbonnier, Chague, Kolo, &
37 Ladermann, 2015), repeated powerful forehand drives may also be a cause of shoulder
38 problems.

39 The achievement of a forehand drive requires three successive phases: a slow backswing to
40 place the body at an optimal pose before the drive, a rapid forward swing to accelerate the
41 racket head for impact, and a follow-through to slowdown the upper limb-racket complex
42 (Ryu, McCormick, Jobe, Moynes, & Antonelli, 1988). Two techniques of forehand drives are
43 mainly used by advanced tennis players: the flat and topspin. The main difference between the
44 two techniques is the racket path during the forward swing. For the flat forehand drive, the
45 racket is mainly moved horizontally, while, for the topspin shot, the racket is first moved
46 downward, then upward until ball impact. Such differences generate a higher vertical velocity
47 of the racket head at impact for topspin than for flat forehand drives (Takahashi, Elliott, &
48 Noffal, 1996). Nevertheless, whatever the technique used and according to the kinetic chain
49 concept described by Elliott and Marsh (1989), the racket head velocity at impact is a result of
50 the cumulative force production at successive body joints. In this context, shoulder motions
51 have a high contribution in the total force production. Takahashi et al. (1996) indeed observed
52 that the shoulder complex accounts for 60% of the upper limb contribution to the racket head
53 velocity during the forward swing phase. The role of the shoulder seems to remain similar

whatever the forehand drive technique, i.e. flat, topspin or lob (Takahashi et al., 1996), the shot velocity (Seeley, Funk, Denning, Hager, & Hopkins, 2011), or the skill level (Landlinger, Lindinger, Stoggl, Wagner, & Muller, 2010).

The primary contribution of the shoulder joint in the racket velocity generation is at the expense of the loadings undergone by this joint. Indeed, the great shoulder joint amplitude and velocity required during the forward swing and follow through phases, i.e. humeral horizontal adduction and internal rotation (Marshall & Elliott, 2000), may lead to high shoulder contact forces. To our knowledge, few studies investigated the loading undergone by the shoulder during forehand drives. Two studies investigated the shoulder loadings by computing the net joint moment/power at the humero-thoracic joint (Bahamonde & Knudson, 2003; Creveaux, Dumas, Cheze, Mace, & Rogowski, 2013a). Bahamonde and Knudson (2003) observed that the horizontal adduction component was the greatest shoulder moment during the forehand drive. Another study investigated the consequences of shoulder loadings during forehand drive by estimating the distance between different shoulder structures (Ladermann, Chague, Kolo, & Charbonnier, 2016). They observed a decrease in the humero-glenoid distance that led, in 29% of the cases, to an anterior-superior humero-glenoid impingement; namely an impingement of the deep surface of the subscapularis tendon and the reflection pulley at the anterior-superior part of the glenoid cavity (Gerber & Sebesta, 2000). Although these studies give a first estimation of the shoulder loadings during the forehand drive, few information are available on the contact forces at the glenohumeral joint.

Contact forces take into consideration both the external forces and the muscle forces acting on the joint (Steele, Demers, Schwartz, & Delp, 2012). Contact forces are usually hard to estimate by experimental assessments since based on invasive methods such as instrumented prosthesis (Nikooyan et al., 2010). Musculoskeletal models therefore offer a promising alternative. Indeed, they make it possible to estimate muscle forces; then the later can be used

in addition to external forces to compute contact forces (Prinold, Masjedi, Johnson, & Bull, 2013; Quental, Folgado, Ambrosio, & Monteiro, 2015). Consequently, musculoskeletal models have been already used to estimate glenohumeral contact forces during wheelchair propulsion (Holloway et al., 2015; van Drongelen, van der Woude, & Veeger, 2011; Veeger, Rozendaal, & van der Helm, 2002), driving task (Pandis, Prinold, & Bull, 2015) or daily life activities (Anglin, Wyss, & Pichora, 2000). Consequently, musculoskeletal model may be helpful to better assess glenohumeral contact forces during tennis forehand drives.

The aim of this study was to investigate the contact force at the glenohumeral joint during the flat and topspin forehand drives in advanced tennis players. We hypothesised that contact force magnitude and orientation would be similar for the flat and topspin forehand drives, and different between the three phases of the stroke.

Methods

Participants

After given their informed consent, 11 right-handed male competitive tennis players (Mean $\pm$ SD age: 29.1 ± 7.6 years; mass: 69.4 ± 5.6 kg; height: 178 ± 6 cm; tennis experience: 22 ± 7 years; weekly training exposure: 5.1 ± 5.8 h; International Tennis Number: 3) volunteered to participate in this study, which was approved by French Ethics Committee Sud-Est II. The players declared no history of injury in the six months before the study. Six players used a semi-western grip, while the other five a western grip, to hit forehand drive. All players were classified in the advanced level (level 3) according to the International Tennis Number (ITN) classification. This meant that the players were able to hit strong forehand drive with control, depth and spin, and to vary spin. The mass, centre of gravity, swingweight and twistweight, i.e. the moments of inertia around the transversal and perpendicular axis of the racket head plane of each player were measured using a Racket Diagnostic Center (Babolat VS, Lyon,

France). Polar moment was calculated by subtracting the twistweight and swingweight (Brody, 1985) (see Figure S1 in the supplementary file for axis definitions).

Instrumentation and data collection

In line with previous study (Rogowski, Creveaux, Cheze, Mace, & Dumas, 2014), each participant was fitted with 17 reflective markers located on the xiphoid process; incisura jugularis; 7th cervical vertebra; 8th thoracic vertebra; and on the right side, angulus acromialis; acromio-clavicular joint; scapula spine (at 1/3 of the spine from the acromion); inferior and superior scapula angles; distal insertion of the middle deltoid; medial and lateral humeral epicondyles; olecranon; radial and ulnar styloid processes; and 2nd and 5th metacarpal heads. In addition, three reflective markers were stuck on the racket and a reflective tape surrounded the ball. The 3D trajectories of the reflective markers were recorded using an optical motion capture system (Motion Analysis, Motion Analysis Corporation, Santa Rosa, CA, USA) comprising eight digital cameras (sampling frequency: 500 Hz).

Procedure

After a 15-min warm-up, each player performed randomly two series of two sets of six successive crosscourt forehand drives; flat forehand drives were hit during one series, and topspin forehand drives during the second one. The players used their own racket (mass: 334 ± 13 g; centre of gravity measured from the handle racket extremity: 328 ± 8 mm; swingweight: 327 ± 9 g.cm²; twistweight: 343 ± 10 g.cm²; polar moment: 15.7 ± 1 g.cm²). They were instructed to hit the ball at their own playing velocity in a comfortable way, when using a square stance. All trials were conducted in an indoor acrylic tennis court, which was instrumented with a ball machine (ball velocity: 16.7 m/s; Airmatic 104, Pop-Lob, Bagneux,

France) located on the baseline of the opposite tennis court, and with a radar gun (Stalker Pro II, Stalker Radar, Plano, TX, USA) located behind the player to measure ball velocity after the ball impact.

Musculoskeletal model

The Delft Shoulder and Elbow model (van der Helm, 1994) available on OpenSim (Delp et al., 2007) was used. Briefly, the model was composed of eight segments (ground, thorax, right clavicle, scapula, humerus, ulna, radius and hand). A racket was added to the model to reproduce the forehand drives. The segments were linked by 18 degrees of freedom namely the ground-thorax joint (three translations and three rotations), sterno-clavicular joint (two rotations), acromio-clavicular joint (three rotations), glenohumeral joint (three rotations), elbow (two rotations) and wrist (two rotations). No degree of freedom was allowed between the hand and the racket. The later was placed in the model for each participant according to its position recorded at the beginning of the forehand drives. The 18 degrees of freedom were actuated by 139 muscle lines of action (Figure 1).

Scaling and kinematics

The generic model was scaled for each participant using the position of the reflective markers recorded during the static pose. In addition, the inertia, mass and size of the participants' racket were used for each scaled model. Then, an inverse kinematic procedure was used to obtain the joint angles by minimising the Euclidian distance between the position of the virtual markers and the position of the experimental markers at each sample time. Moreover, an ellipsoid fitted to the semi-right-thorax was implemented with a two-point-on-ellipsoid constraint ensuring the scapula to follow the thoracic curvature throughout the movement (Michaud, 2016, July). For further analysis, joint angles were filtered using a zero-lag fourth-

order low-pass Butterworth filter with an 8 Hz cut-off frequency. To avoid problems of data filtering through ball impact (Knudson & Bahamonde, 2001), three frames around the ball impact (determined according to Creveaux et al. (2013b)) were deleted; then the data were filtered separately before and after the ball impact.

Glenohumeral contact forces

The joint kinematics and scaled models were used as input of the static optimisation procedure (Delp et al., 2007) in order to obtain muscle activations required to perform the forehand drives. Briefly, at each sample time (t), the objective function (J) to minimise consisted in the quadratic sum of muscle activations (a) and the quadratic sum of residual torques (i.e. artificial joint actuators used to solve the numerical problem when muscles cannot reproduce the joint torque computed with inverse dynamics procedure) (τ^{res}) at each degree of freedom (n), subject to two constraints 1) the joint torques calculated from muscle activations were equal to the torques (τ) computed from the inverse dynamic procedure (Eq. 2) and 2) the glenohumeral forces were inside a friction polygon ensuring the stability of the glenohumeral joint (Eq. 3) (Dickerson, Chaffin, & Hughes, 2007).

$$\min_{a(t)_m} J(t) = c_1 * \sum_{m=1}^{139} (a(t)_m)^2 + c_2 * \sum_{n=1}^{18} (\tau^{res}(t)_n)^2 , \quad (\text{Eq. 1})$$

subject to

$$\sum_{m=1}^{139} r_{m,j} \cdot [a(t)_m \cdot f(F_m^0, l_m, v_m)] + \sum_{n=1}^{18} \tau^{res}(t)_n = \tau(t)_j , \quad (\text{Eq. 2})$$

$$\begin{bmatrix} \alpha_1\mu_1 & -1 & 0 \\ \alpha_2\mu_2 & -1 & -1 \\ \alpha_3\mu_3 & 0 & -1 \\ \alpha_4\mu_4 & 1 & -1 \\ \alpha_5\mu_5 & 1 & 0 \\ \alpha_6\mu_6 & 1 & 1 \\ \alpha_7\mu_7 & 0 & 1 \\ \alpha_8\mu_8 & -1 & 1 \end{bmatrix} \cdot \begin{bmatrix} R_{GH}^n \\ R_{GH}^{s1} \\ R_{GH}^{s2} \end{bmatrix} \geq \mathbf{0} \quad , \quad (\text{Eq. 3})$$

172 with m referring to the 139 muscle-tendon-units; F_m^0 is the maximum isometric force of the
 173 modelled muscles m ; l_m is the muscle model length; v_m denotes the muscle model
 174 shortening/lengthening velocity; $f(F_m^0, l_m, v_m)$ is its force-length-velocity surface
 175 relationship; $r_{m,j}$ is the moment arm about j^{th} joint axis; c_1 and c_2 referring to weighting
 176 factors; $R_{GH}^n, R_{GH}^{s1}, R_{GH}^{s2}$: the normal, posterior-anterior and superior-inferior glenohumeral
 177 contact force components; μ_i : the dislocation thresholds equal to [0.29, 0.33, 0.51, 0.35, 0.30,
 178 0.43, 0.56, 0.4] (Dickerson et al., 2007) for each polygon vertex (i), α_i : optimised coefficients
 179 (Wieber, 2001) to best match with Dickerson's polygon (2007).

180 The glenohumeral contact forces between the humeral head and scapula $\vec{R}_{GH}$ were
 181 calculated using the *jointReaction* algorithm of OpenSim 3.2 (Delp et al., 2007; Steele et al.,
 182 2012):

$$\vec{R}_{GH} = [M]_{humeralus} \cdot \vec{a}_{humeralus} - \left(\vec{R}_{elbow} + \sum \vec{F}_{muscles} + \vec{F}_{gravity} \right), \quad (\text{Eq. 4})$$

183 $[M]_{humeralus}$ is the matrix of inertial properties of the humerus, $\vec{a}_{humeralus}$ denotes the six
 184 dimensional angular and linear accelerations of the humerus, $\vec{R}_{elbow}$ is the force and moment
 185 of the ulna on the humerus; $\vec{F}_{muscles}$ and $\vec{F}_{gravity}$ are the muscle and gravitational forces and
 186 moments acting on the humerus. Glenohumeral contact forces were initially calculated in the
 187 scapula reference system. In order to express these forces in the glenoid reference system,
 188 three landmarks were located on the scapula of the generic Opensim model (most anterior,
 189 posterior and inferior points of the edge of the glenoid). These landmarks were used to define

the reference system of the glenoid. Finally, the transformation matrix scapula-to-glenoid was used to compute glenohumeral contact forces in the glenoid reference system.

Analysis

Among all forehand drives performed, only the three for which the ball bounced in the opposite court and with similar ball post-impact velocities were taken into consideration in the subsequent analysis. Four key-events were detected, such as the beginning of the shoulder turn, first forward movement of the racket, ball impact, and highest elbow height after impact (Ryu et al., 1988), in order to define three successive phases: (1) the backswing; (2) the forward swing and (3) the follow-through.

For each phase, the mean and maximal values of the compressive, shearing and total glenohumeral contact forces were calculated. In line with a previous study (Charbonnier et al., 2015), the direction of the glenohumeral contact forces was expressed by dividing the glenoid into eight sections (Figure 2). Then, the percentage of glenohumeral contact forces inside each of the eight sections was computed. Finally, for each variable cited above, the values obtained either for the flat or topspin forehand drives were averaged separately, in order to obtain one value per variable and per forehand drive technique.

In order to evaluate the model outputs, mean and maximal residual torques were computed and compared to the joint torques calculated by the inverse dynamic procedure. In addition, the mean activation of nine muscles (anterior, middle, posterior deltoids; pectoralis major; latissimus dorsi; serratus anterior; infraspinatus; supraspinatus and subscapularis) was calculated and ranked by order of intensity to be compared with EMG data from the literature. The rank comparison was chosen instead of the comparison of activation values since musculoskeletal model activations are different from experimental normalised EMG.

Statistical analysis

The interaction effect between the forehand drive technique (flat vs. topspin; fixed effect) and the phase (backswing vs. forward swing vs. follow-through; fixed effect) on the mean and maximal compressive, shearing and total glenohumeral contact forces was tested using linear mixed models. Linear mixed-model is an alternative method to the ANOVA on repeated measures that may be more advantageous especially with categorical data (Barr, Levy, Scheepers, & Tily, 2013). As within-participants repeated measures were performed, participants were entered as random intercept. To avoid the effect of a different racket velocity at the frame before the impact between the flat and topspin forehand drives, the racket velocity was also entered as random intercept. Finally, as the forehand drive technique might have different effect with respect to the participants and/or the racket velocity, forehand drive technique was also entered as random slope (Barr et al., 2013). The p -values were obtained by likelihood ratio tests of the full model against the model without the effect in question. The level of significance was set at $p < 0.008$ (Bonferroni correction $0.05/6$ variables). When a main effect was revealed by the linear mixed-model, post-hoc pairwise comparisons were performed using the Tukey's honestly significant difference procedure with a significant p -value set at 0.05. The linearity, homoscedasticity and normality of the linear-mixed model residuals were graphically controlled. All models were performed using the package *lme4* of R software (R 3.2, RCore Team 2014, Vienna, Austria).

The chi-square test of independence was used to assess if the proportion of glenohumeral contact forces inside the eight glenoid sections was dependent of the forehand drive technique. Finally, the difference in the proportion of glenohumeral contact forces between each glenoid section was tested using the chi-square test for homogeneity of proportion. All chi-square tests were executed using R software.

Results

Evaluation of the model outputs

Regardless the forehand drive technique, the maximal residual torques were on average 14.7 ± 2.3 , 8.4 ± 4.9 and 22.6 ± 7.2 times smaller than the corresponding maximal joint torques for the sterno-clavicular, acromio-clavicular and glenohumeral joint respectively. Greater differences were observed concerning the mean residual torques which were in average 29.7 ± 7.1 , 14.3 ± 6.8 and 37.6 ± 10.5 times smaller than the corresponding mean joint torques for the sterno-clavicular, acromio-clavicular and glenohumeral joint, respectively.

The qualitative analysis of the muscle activations enabled to point out low activations for all muscles during the backswing phase (Figure 3). The pectoralis major/minor, anterior deltoid, subscapularis and upper trapezius were highly activated in comparison to the other shoulder muscles during the forward swing phase (Figure 3). Finally, the middle and posterior deltoids, latissimus dorsi, infraspinatus, upper and middle trapezius were mainly involved during the follow-through phase. The ranking order of muscle mean activation between phases was overall similar to those reported by Ryu et al. (1988) (Table 1).

Forehand drive analysis

The mean ball velocities were 31.5 ± 5.5 m/s for flat and 27.4 ± 4.6 m/s for the topspin forehand drives. The norms of the 3D racket velocity one frame before the impact were on average 20.2 ± 4.5 m/s for flat and 21.7 ± 5.3 m/s for the topspin forehand drives. Finally, the mean racket vertical velocities were 6.6 ± 2.1 m/s for flat and 11.1 ± 1.9 m/s for the topspin forehand drives.

The linear mixed models revealed only a main effect of the phase on the mean and maximal compressive, shearing and total glenohumeral contact forces ($\chi^2(2) \in [68.7, 87]$, $p < 0.001$). The glenohumeral contact forces during the forward swing phase were in average 1.25 time

greater than during the follow-through phase, and 5.8 times greater than during the backswing phase (Table 2 and figure 4).

Concerning the repartition of the glenohumeral contact forces in the eight glenoid sections, the chi-square test of independence revealed no dependency between the forehand drive technique and the glenoid sections either for the backswing ($\chi^2(7)=0.34$, $p=0.99$), forward swing ($\chi^2(7)=2.41$, $p=0.93$) or follow-through phases ($\chi^2(7)=8.59$, $p=0.28$). Regardless the forehand drive technique, the chi-square test for homogeneity revealed for each phase that the distribution of the glenohumeral contact forces in the sections of the glenoid was not homogeneous ($\chi^2(7)=462$, $p<0.001$; $\chi^2(7)=440$, $p<0.001$; $\chi^2(7)=167$, $p<0.001$, respectively for the backswing, forward swing and follow-through phases). For the backswing and forward swing phases, the greatest residual of the chi-square tests was observed for the “anteriorSuperior” section (Figure 2), while concerning the follow-through phase, the greatest residual of the chi-square tests was observed for the “superiorAnterior” section (see Table T1 of the residuals in the supplementary file).

Discussion and implications

This study aimed to investigate the contact forces at the glenohumeral joint during the flat and topspin forehand drives in advanced tennis players. The first findings were that the forehand drive technique, i.e. flat or topspin, did not change the glenohumeral contact forces undergone by the tennis player. The second findings were that the maximal contact force undergone by the glenohumeral joint occurred during the forward swing phase, and pointed toward the anterior-superior part of the glenoid.

Evaluation of the model outputs

As the computed glenohumeral contact forces resulted from the modelled muscle forces, the validity of our model needed to be evaluated. The values for the maximal and mean residual torques involved during the forehand drive motion were between 8.4 and 37.6 times smaller than the maximal and mean torques computed with the inverse dynamic procedure. These results meant that the muscles included in the model were sufficient to ensure the joint torques necessary to perform the forehand drives (Hicks, Uchida, Seth, Rajagopal, & Delp, 2015). Regardless the forehand drive technique, the orders of the magnitude of muscle activations between phases was similar than the normalised EMG data of Ryu et al. (1988). Only three muscles had a different ranking, namely the infraspinatus, supraspinatus and middle deltoid muscles (Table 1). The mean normalised EMG levels of these muscles however present small differences between the phases of the forehand drive, i.e. 3 to 11% (Ryu et al., 1988), potentially explaining the discrepancies in the rank between studies. The changes in racket specifications over the last 30 years may also result in differences in player's motion, and then in muscle demand, hence explaining some of the discrepancies between the two studies. In comparison with Genevois, Creveaux, Hautier, and Rogowski (2015) and Rota, Morel, Saboul, Rogowski, and Hautier (2014), we also observed that the mean activations of the anterior deltoid and pectoralis major were greater than the mean activations of the latissimus dorsi and middle deltoid (Figure 3).

In summary, our model seemed to respect the salient characteristic of the forehand drive performed by the players. Although our computed glenohumeral contact forces could not be compared with in-vivo data (see limitations section), we assume that they were suitable for further interpretation.

Glenohumeral contact forces

The contribution of the shoulder complex to the racket velocity production remains similar whatever the technique required to hit a forehand drive (Takahashi et al., 1996). This outcome may partly explain the lack of differences in the glenohumeral contact forces computed during both the flat and topspin forehand drives. It is also possible that our model was not sufficiently accurate to detect the small changes in glenohumeral contact forces with respect to the forehand drive technique. Grouping the results for both techniques of shots nevertheless provided new information on the glenohumeral contact forces during the tennis forehand drive.

The achievement of a forehand drive requires firstly a slow motion, then a rapid acceleration, and finally a deceleration of the upper limb – racket complex (Elliott & Marsh, 1989). Such a motion pattern may explain the differences observed in maximal glenohumeral contact forces between the three phases, e.g. 559 ± 145 N, 3573 ± 1383 N and 2327 ± 1162 N, for the backswing, forward swing and follow-through phase respectively. These results suggested that the player's shoulder was not at the same level of risk for injury throughout the forehand drive motion. During the backswing phase, a trunk rotation occurs to place it perpendicularly to the net (Elliott & Marsh, 1989), while the humerus abducts, extends, and externally rotates to take back the racket. The activation of the teres minor muscle may contribute to place the humerus in the adequate position at the end of the backswing, whereas the pectoralis minor muscle mainly acts to anteriorly tilt the scapula (Rogowski, Creveaux, Cheze, & Dumas, 2014) (Figure 3). The anterior and superior orientation of the glenohumeral contact forces may be partly related to the early activation of the anterior deltoid muscle, which is involved in the stretch-shorten cycle movement to contribute to a powerful shot (Wilson, Elliott, & Wood, 1991). As the positioning of the racket at the end of the backswing phase requires low muscle activations, the shearing forces and consequently the compressive forces (that ensure the glenohumeral stability) remained at low levels. During the forward swing phase, a large

demand is placed on the pectoralis major/minor, anterior deltoid and subscapularis muscles to achieve a rapid humeral horizontal adduction and internal rotation (Figure 3) (Ryu et al., 1988). As these muscles cross the glenohumeral joint, their increased activation automatically led to raise the compressive and shearing glenohumeral contact forces and to orientate them toward the anterior section of the glenoid. It may be hypothesised that the activation of the latissimus dorsi, teres minor, infraspinatus, and rhomboid became crucial to counteract the increased shearing glenohumeral force. Their activation may have contributed to the increased compressive glenohumeral force, ensuring then the glenohumeral stability (Dickerson et al., 2007; Lippitt & Matsen, 1993). During the follow-through phase, the glenohumeral contact forces decreased, even if they remained relatively high in comparison to the backswing phase. This may be explained by the high velocity of the arm during the follow-through phase (Marshall & Elliott, 2000). The middle deltoid and upper trapezius were involved in the humeral abduction, and may contribute to upwardly orientate the glenohumeral contact forces. The middle trapezius, infraspinatus, teres minor and posterior deltoid muscles then may contract eccentrically to resist to the anterior glenohumeral shearing forces. Although our model did not allow translations of the humeral head, the observation of the orientation of the glenohumeral contact forces leads to the same conclusions than Ladermann et al. (2016). The muscle forces produced during the forehand drive result in glenohumeral contact forces pointing in a forward and upward direction; the consequence may be a translation of the humeral head that may cause an anterior-superior humero-glenoid impingement.

Limitations

The first limitation of this study was that the glenohumeral contact forces were estimated and not measured in-vivo. Nevertheless, only invasive methods make possible to measure glenohumeral contact forces. To the best of our knowledge, few studies measured in-vivo

glenohumeral contact forces, only focusing on arm elevation, isometric forces (Nikooyan et al., 2010) and daily living activities (Westerhoff et al., 2009, 2012), but not during sport movements. In addition, to our knowledge, glenohumeral contact forces estimated through musculoskeletal model were only investigated during wheelchair tasks (Holloway et al., 2015; van der Helm & Veeger, 1996; van Drongelen et al., 2011), but not for sport movements. Although the comparison between our results and the previous studies is not obvious, our computed contact forces (Table 2) were up to four times greater than the maximal glenohumeral forces measured in-vivo with an instrumented prosthesis during arm elevation (≈ 800 N.) (Nikooyan et al., 2010). In comparison to other musculoskeletal models, we found glenohumeral contact forces in the same range than those estimated during wheelchair propulsion (Holloway et al., 2015; van der Helm & Veeger, 1996) or during vertical weight relief lift from a wheelchair (van Drongelen et al., 2011). Consequently, the values of the glenohumeral contact forces obtained in this study must be taken with caution even if they do not seem to be abnormal in comparison to the literature.

The second limitation was that our model enabled only three rotations at the glenohumeral joint and no translation. Although the computing of the glenohumeral contact forces is probably affected by the number of degrees of freedom, the orientation of the glenohumeral contact forces calculated in our study led to similar conclusions than those of Ladermann et al. (2016) who had implemented glenohumeral translation in their kinematic model. In addition, our kinematical model obtained only from skin markers did not enable to estimate accurately the very small glenohumeral translations occurring during forehand drives (less than 2.8 mm for forehand drive mimic) (Dal Maso et al., 2015). The third limitation of our study was that the muscle forces of the model were not specific to those of the participants. It may be possible that the tennis players of this study had different maximal isometric muscle forces than those implemented in the generic Delft Shoulder and Elbow model. Further

studies are needed with participant-specific musculoskeletal model to estimate more precisely
glenohumeral contact forces.

Conclusion

This study was the first to estimate glenohumeral contact forces during forehand drives,
through musculoskeletal modelling. Regardless the forehand drive technique (flat or topspin),
the tennis players underwent the greatest glenohumeral contact forces during the forward
swing, then during the follow-through and finally during the backswing phase. Most of the
glenohumeral contact forces pointed toward the anterior-superior part of the glenoid. This
study brings new knowledge on the glenohumeral loadings during real sport motion, and
provides information that may help the sport clinicians and tennis coaches on the causative
factors for shoulder problems.

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Table 1. Ranked order between phases by intensity of muscle activations or normalised EMG for seven shoulder muscles in comparison to the data of Ryu et al. (1988). The greater the rank, the greater the activation intensity or normalised EMG.

		Backswing	Forward swing	Follow-through
Pectoralis Major	Ryu (1988)	1	3	2
	Our model	1	3	2
Latissimus dorsi	Ryu (1988)	1	2	3
	Our model	1	2	3
Serratus Anterior	Ryu (1988)	1	3	2
	Our model	1	3	2
Middle deltoid	Ryu (1988)	3	1	2
	Our model	1*	2*	3*
Infraspinatus	Ryu (1988)	2	1	3
	Our model	1*	2*	3
Supraspinatus	Ryu (1988)	2	3	1
	Our model	1*	2*	3*
Subscapularis	Ryu (1988)	1	3	2
	Our model	1	3	2

* *denotes a difference in rank order between our model and Ryu's data.*

Table 2. Mean $\pm$ standard deviation of the mean and maximal compressive, shearing and total glenohumeral forces with respect to the phase (backswing, forward swing and follow-through). As no effect of the forehand drive technique has been observed, values for the flat and topspin forehand drives were averaged.

	Backswing	Forward swing	Follow-through	p-values
Maximal				
Compressive (N)	530 $\pm$ 137	3396 $\pm$ 1323	2210 $\pm$ 1103	***, ###, ʒʒʒ
Shear (N)	176 $\pm$ 49	1116 $\pm$ 410	754 $\pm$ 361	***, ###, ʒʒʒ
Total (N)	559 $\pm$ 145	3573 $\pm$ 1383	2327 $\pm$ 1162	***, ###, ʒʒʒ
Mean				
Compressive (N)	262 $\pm$ 65	1513 $\pm$ 643	1215 $\pm$ 596	***, ‡, ʒʒʒ
Shear (N)	87 $\pm$ 23	498 $\pm$ 211	394 $\pm$ 221	***, ‡, ʒʒʒ
Total (N)	276 $\pm$ 69	1593 $\pm$ 677	1281 $\pm$ 633	***, ‡, ʒʒʒ

* indicates a significant difference between the backswing and forward swing phases (* for $p < 0.05$; ** for $p < 0.01$; *** for $p < 0.001$); ‡ indicates a significant difference between the forward swing and follow-through phases (‡ for $p < 0.05$; # for $p < 0.01$; ### for $p < 0.001$); ʒ indicates a significant difference between the backswing and follow-through phases (ʒ for $p < 0.05$; ʒʒ for $p < 0.01$; ʒʒʒ for $p < 0.001$).

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552 **Figure 1.** Delft Shoulder and Elbow model implemented with a tennis racket

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557 **Figure 2.** The glenoid was divided into eight sectors based on Charbonnier et al. (2015) in
558 order to represent the direction of the glenohumeral forces: 1-anteriorSuperior, 2-
559 superiorAnterior, 3-superiorPosterior, 4-posteriorSuperior, 5-posteriorInferior, 6-
560 inferiorPosterior, 7-inferiorAnterior and 8-anteriorInferior

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Figure 3. Muscle activations of a representative participant obtained with the musculoskeletal model. Zero second fits the ball impact. The vertical dashed and solid black lines correspond to the beginning of the forward swing phase for the flat and topspin forehand drives respectively.

Figure 4. Compressive (dashed line), shearing (solid line) and total (dotted line) glenohumeral (GH) forces during the flat and topspin forehand drives of a representative participant with respect to the time. Zero second fits the ball impact.

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Figure 5. Mean $\pm$ standard deviation distribution (%) of the glenohumeral contact forces

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during the forehand drives with respect to the stroke phase. As no dependency between the

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forehand drive technique and the glenoid section was observed, values for the flat and topspin

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forehand drives were averaged.

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