The effect of intracortical bone pin application on kinetics and tibiocalcaneal kinematics of walking gait

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Abstract

Bone anchored markers using intracortical bone pins are one of the few available methods for analyzing skeletal motion during human gait in-vivo without errors induced by soft tissue artifacts. However, bone anchored markers require local anesthesia and may alter the motor control and motor output during gait. The purpose of this study was to examine the effect of local anesthesia and the use of bone anchored markers on typical gait analysis variables. Five subjects were analyzed in two different gait analysis sessions. In the first session, a protocol with skin markers was used. In the second session, bone anchored markers were added after local anesthesia was applied. For both sessions, three dimensional infrared kinematics of the calcaneus and tibia segments, ground reaction forces, and plantar pressure data were collected. 95% confidence intervals and boxplots were used to compare protocols and assess the data distribution and data variability for each subject. Although considerable variation was found between subjects, within-subject comparison of the two protocols revealed non-systematic effects on the target variables. Two of the five subjects walked at reduced gait speed during the bone pin session, which explained the between-session differences found in kinetic and kinematic variables. The remaining three subjects did not systematically alter their gait pattern between the two sessions. Results support the hypothesis that local anesthesia and the presence of bone pins still allow a valid gait pattern to be analyzed.

Keywords

Introduction

Over the past decades, marker-based optoelectronic stereophotogrammetry has become a standard procedure in many types of dynamic analyses of the lower extremity and the foot during human

locomotion. However, the limitations of skin based marker placement (SKIN), especially the presence of soft tissue artifacts, are well-known [1]. Bone anchored markers using intracortical bone pins (PIN) are considered to be one of the few available methods for analyzing skeletal motion of the lower extremity in-vivo without errors induced by soft tissue artifacts. Another appealing advantage of PIN over SKIN is the fact that bony structures of the foot and ankle can be analyzed that are either too small for valid marker placement, suffer from heavy skin movement artifacts, or are inaccessible by surface markers, e. g. the navicular or the talus. Other techniques of image-based kinematic analyses, like magnetic resonance imaging, radiostereometric analysis or fluoroscopy, are currently only capable of capturing static or slow dynamic situations and thus may not be feasible for analyzing human locomotion.

The most evident drawback of PIN is the requirement of local anesthesia and its potential effects on motor control and motor output. Previous studies have demonstrated effects of local anesthesia on roll-over during gait, e. g. the increase of metatarsal head loading in plantar pressure data after applying ice to the plantar surface of the foot [2]. Furthermore, the mechanical intervention of protruding pins anchored to the bone might interfere with the motion of tendons, ligaments, skin and other pins applied in close proximity. Psychological effects of PIN cannot be ruled out either, as subjects may alter their gait patterns towards a more cautious roll-over pattern and less range of motion.

This study aimed to analyze whether the application of PIN results in altered motor output and thus, whether PIN is a valid method for analyzing human gait. The validity of PIN can only be assured if the magnitude and variability of human gait kinematics and kinetics are unaffected by local anesthesia and pin application.

An experimental protocol was designed with PIN as an independent variable, and with kinematics and kinetics collected as target variables. It was hypothesized that target variable magnitude and variability would not differ between use of PIN and SKIN protocols.

The outcome of this study is an important step to validate previous and future studies, in which a

similar PIN approach is used to analyze kinetics and kinematics during human gait [3–10].

Methods

Five healthy male subjects participated in the study. Anthropometric data for each subject are listed in Table 1. A sixth subject had to be excluded due to problems with the application of the pins. Study procedures were approved by the local ethics committee of the Karolinska Institute, Huddinge, Sweden.

Insert Table 1 here

Subjects were analyzed during normal walking gait in laboratory conditions, collecting kinetics and kinematics of their movement patterns. Two measurement sessions were carried out for each subject in the following sequence, either on the same day or on two consecutive days: during session 1 (SKIN), a marker set of skin-mounted markers was applied to the calcaneus and tibia of each subject. Session 2 (PIN) succeeded SKIN and was conducted after local anesthetics had been applied and intracortical bone pins had been inserted to the bony structures of the subjects' right foot. Hence, a combined marker set of SKIN and PIN markers was used in session 2 (see Figure 1). The locations of SKIN markers were marked directly on subjects' skin to ensure the reproducible placement of SKIN markers after pin insertion.

Insert Figure 1 here

Each subject was asked to complete 9-12 SKIN and at least 20 PIN walking gait trials. Data were collected while subjects walked on a 9.5 m walkway at their self-selected pace.

Kinematics were recorded using a 12-camera infrared system (Qualisys, Gothenburg, Sweden)

sampling at 240 Hz. Kinetics were captured by a force plate (Kistler, Winterthur, Switzerland, 960 Hz, sync to Kinematics) and a pressure distribution platform (Emed, Novel, Munich, Germany, 50 Hz), which were mounted flush with the walkway and approximately 1 m apart from each other in order to capture two consecutive foot strikes of the same foot.

Differences between SKIN and PIN were assessed by analyzing relative motion of the shank and heel. In both conditions, four markers were attached to the calcaneus (one lateral, one medial, two posterior) in such a way as to prevent artifacts from the motion of the calcaneal fat pad. The motion of the shank was captured using six markers (fibular head, tibial tuberosity, middle of tibia crest, both malleoli, and the lower lateral malleolus, see [11] for marker positions). Gaps in marker trajectories were interpolated up to a gap size of 10 frames (42ms) using a higher order polynomial fit. Trials featuring larger trajectory gaps were discarded.

The technical reference frame for each segment was set up during relaxed standing in a neutral foot position, i.e., foot length axis was in parallel to the sagittal plane. The segment origin was placed at the mean marker location of all segment markers, and axes were aligned parallel to the global reference frame. Segment motions were calculated by matching corresponding markers across frames according to the procedures described by [12]. This approach is robust against small deviations of marker placement between sessions, and does not rely on any markers placed on specific anatomical locations. Relative motion of the heel versus the shank in anatomical planes was calculated by projecting the rotation of the heel around its helical axis onto the shank coordinate system, accounting for the instantaneous helical axis orientation.

The range of motion (ROM) in three anatomical planes during the stance phase of subjects' gait was calculated as the target variable in order to allow for simple interpretation and comparison of the obtained results to other studies [3,11,13]. The stance phase was determined by setting a 10 N threshold on the Kistler force plate. Since the EMED platform was not synchronized to the kinematic system, the second stance phase was detected based on a local extremum of the acceleration of the heel markers similar to an approach based on foot velocity described by [14].

Two local maxima (Fz2 and Fz4) and one local minimum (Fz3) [15,16], as well as their relative timing (Fz2_t, Fz3_t, Fz4_t), and ground contact time (GCT) were calculated from force plate data. Gait speed was calculated based on the distance between two consecutive foot strikes, which was determined from the average positions of all heel markers. Kinematic data and force plate data were processed using custom-written Matlab software (The Math Works, Natick, USA).

Plantar pressures were recorded using the EMED AT platform (4 sensors/cm², 50 Hz). Raw data were exported to ASCII and analyzed using custom-written Matlab software as described in [17], including the definition of anatomical regions (masks M1 through M9, see Figure 3). For the analyses conducted in this study, the relative force-time integral (FTI) of each mask as well as the timing of heel-off during roll-over (HO) were calculated. FTI was calculated as relative impulse of each mask compared to the total impulse of the foot area. HO was determined as the relative point in stance phase, at which the cumulated force of the heel sensors dropped below 10% of the total maximum force recorded during the entire stance phase. Other studies have employed similar variables to analyze roll-over timing based on plantar pressure data [18].

Arch regions M4 and M5 were not considered for analysis in this study, as none of the subjects' feet had any deformity or pathology, and thus most of the loads in M4 and M5 were near or exactly zero for each stance phase.

Insert Figure 2 here

Due to the small sample size and the explorative nature of this study, we did not conduct any statistical testing. Instead, 95%-confidence intervals (upper bound: up95, lower bound: lo95) for the mean of each variable and condition were calculated from sample data. Data distribution and variance were assessed graphically using boxplots.

Non-overlapping confidence bounds for SKIN and PIN indicate possible differences between the two conditions. The practical relevance of this difference was rated based on the repeatability of the

respective variable, which– due to the small sample of our dataset – was synthesized from other studies involving larger samples. As long as differences between SKIN and PIN did not exceed the expected random fluctuations of repeated measurements of a variable, they were considered practically irrelevant [19].

Results

Technical problems caused missing values for some of the subjects (kinematics: trajectory gaps in S3, GRF: operator faults in S2 & S4). Table 2 contains the 95%-confidence bounds for all variables. Non-overlapping bounds are shaded grey. Figure 3 depicts the distribution of GRF data as an example.

Insert Table 2 here

None of the collected variables indicated systematic differences between SKIN and PIN conditions across all subjects. Differences in GRF were aggregated for S1, and kinematic differences aggregated for S5. These cases will be discussed in the following section. All other differences appeared to occur randomly and in a non-systematic manner.

Insert Figure 3 here

Discussion

The aim of this study was to investigate the influence of intracortical bone pins on the kinematics and kinetics of walking gait. GRF, plantar pressures, and 3D kinematics were measured with and without intracortical bone pins applied to the subjects' foot and shank, and compared between conditions. We hypothesized that the presence of pins would not systematically alter subjects' gait patterns. Due to the invasive nature of bone pin application, only a small sample of five subjects

was available for analysis.

The results of these five subjects do not indicate a systematic effect of intracortical bone pins on the gait pattern during walking. Subjects 1 and 5 effectively reduced their gait speed after pin application, which explains the reduced excursion of vertical GRF curves, especially at Fz4. The pattern of GRF changes observed in our data was similar across subjects (lower Fz2 and Fz4 & larger Fz3 and GCT for PIN). This can be interpreted as a less pronounced foot strike and a more cautious stance phase in the PIN condition and may be directly linked to gait speed [20]. Relative timing of GRF variables was only marginally influenced by PIN.

Effects of PIN on kinematic data appear to be less uniform. Only the data of S5 indicate an increase of transversal and a decrease of sagittal ROM in the PIN condition. The remaining subjects' data indicate a substantial amount of variability across trials even within conditions. However, similar effects have been observed in other studies [11] and may mask smaller, but more systematic effects. Being smaller than the normal variation we assume these are incidental and do not compromise the validity of the kinematic data.

Plantar pressure distributions did not indicate any systematic effect between SKIN and PIN. Only the increase of medial forefoot loading in S3 corresponded to the interpretation of GRF data that subjects tend to demonstrate a more cautious gait pattern in PIN.

In summary, PIN induced a systematically decreased gait dynamic in one of our subjects (S1). Across all subjects, effects on kinetics were consistently lower than 5% (20-30 N vertical GRF, 1% ROP in timing variables, 1% of relative FTI). These effect magnitudes were considered subtle, compared to the random fluctuations expected from repeated measures in this type of experimental design [21,22]. The most prominent differences between SKIN and PIN occurred in kinematic data with up to 25% of ROM (~2-3°). In light of the more variable nature of kinematic data, the SKIN-PIN differences in our data were considered small, based on the amount of variation to be expected from repeated trials collected with a similar protocol, but with only skin markers attached [11]. Therefore, we considered the observed differences in kinetics and kinematics to be practically

irrelevant given the limiting factors of experimental protocols in laboratory gait analysis.

The size of the 95%-confidence-intervals as well as the size of the box-plots indicate that SKIN and PIN conditions caused the same amount of variability in the recorded data, for most of the subjects and most of the collected variables. Based on this observation, we did not find evidence for a less reliable, less secure or less stable gait in the PIN condition.

Therefore, the data of our study support our hypothesis that there is no difference in gait between SKIN and PIN conditions. A similar result was already observed for slow running gait by [3]. Both studies support the assumption that intracortical bone pins and local anesthesia still allow for a valid gait pattern to be analyzed.

Moreover, the authors think that the increasing availability of studies on the reliability of classic gait analyses causes a more general problem with these kinds of protocols: biomechanical data collected during human locomotion contains substantial amounts of variability across repeated trials, which statistically override the smaller systematic effects that researchers aim to observe when they conduct studies involving gait analyses. This phenomenon does not appear to be causally linked to invasive bone pin application. Gait variability appears to be a more generic feature of human motor performance, and becomes especially problematic when gait analysis protocols are conducted in a laboratory environment with a limited number of foot strikes to be captured and analyzed [23].

In the past, studies involving intracortical bone pins have not been conducted with sample sizes that would allow the generalization of the observed effects. Since a non-random sample was recruited in our study, its scope is of a predominantly descriptive nature. External validity may seem limited with such protocols. However, the benefit of applying invasive bone pins can be found in their contribution to the inductive reconstruction of theories about the kinematic mechanisms of bony structures, e. g. the amount of bony movement within the upper and lower ankle joints during gait [3]. Results of bone pin studies may be unexpected at first and specific to a small group of individuals, but may in turn stimulate the evolution of theories.

Conclusion

The results of this study suggest that intracortical bone pins applied under local anesthesia do not alter subjects' gait patterns in a systematic way. Data of gait analysis protocols using intracortical bone pins should therefore be comparable to protocols using skin mounted markers.

Acknowledgements

Conflict of interest statement

The authors report no conflicts of interest.

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Table 1: Anthropometric data for all five subjects

Subject	Age [y]	Height [m]	Weight [kg]	BMI [kg*m^{-2}]
S1	36	1.80	70	21.6
S2	57	1.83	94	28.1
S3	35	1.73	75	25.1
S4	38	1.82	112	33.8
S5	32	1.80	71	21.9
Mean (SD)	39.6 (10.0)	1.80 (0.04)	84.4 (18.2)	26.1 (5.1)

Table 2: Overview of GRF, kinematics, and plantar pressure data of all five subjects

			S1 skin	S1 pin	S2 skin	S2 pin	S3 skin	S3 pin	S4 skin	S4 pin	S5 skin	S5 pin
GRF		Fz2 [N] up95	765.4	710.0			990.8	931.4			1000.4	962.7
		Fz2 [N] lo95	735.7	672.0			946.8	902.0			938.8	910.5
		Fz2_t [% ROP] up95	20.5	20.2			21.5	21.3			20.6	20.8
		Fz2_t [% ROP] lo95	19.0	19.1			20.5	19.6			19.6	19.6
		Fz3 [N] up95	456.3	477.7			474.5	506.9			437.1	447.4
		Fz3 [N] lo95	440.6	457.0			447.5	490.6			397.8	409.7
		Fz3_t [% ROP] up95	47.0	44.5			50.2	48.1			50.0	48.2
		Fz3_t [% ROP] lo95	44.9	41.8			46.9	46.4			48.2	47.0
		Fz4 [N] up95	740.3	709.1			862.3	843.1			880.0	825.9
		Fz4 [N] lo95	716.1	694.4			832.0	821.7			830.9	808.2
		Fz4_t [% ROP] up95	77.8	75.8			78.1	77.9			77.2	76.2
		Fz4_t [% ROP] lo95	76.6	74.7			76.6	77.2			76.0	75.4
		GCT [s] up95	0.642	0.670			0.630	0.658			0.636	0.628
		GCT [s] lo95	0.624	0.655			0.620	0.640			0.622	0.606
Kinematics	Calcaneus vs. Tibia	ROM frontal [°] up95	11.1	17.4	12.5	13.2			12.1	9.5	8.5	5.7
		ROM frontal [°] lo95	9.5	8.8	10.6	8.9			9.3	8.5	6.3	4.9
		ROM sagittal [°] up95	17.0	21.5	14.9	13.7			14.1	15.1	14.9	11.1
		ROM sagittal [°] lo95	15.3	14.5	12.0	12.4			10.3	13.8	13.1	10.2
		ROM transversal [°] up95	9.5	13.2	9.6	8.8			11.4	9.4	6.8	10.3
		ROM transversal [°] lo95	7.5	9.9	5.5	6.5			4.4	8.6	5.4	9.0
		gait speed [m/s] up95	1.49	1.36	1.42	1.44			1.52	1.49	1.55	1.48
		gait speed [m/s] lo95	1.42	1.28	1.38	1.38			1.45	1.43	1.51	1.45
Pressure Distribution		HO [% ROP] up95	67.9	67.9	75.0	63.5	54.5	53.6	66.2	59.9	53.6	53.7
		HO [% ROP] lo95	64.7	57.6	43.3	55.2	49.4	45.6	59.1	54.8	49.1	50.4
		FTI M1 [%] up95	23.3	24.3	25.5	22.2	20.3	20.8	20.3	19.7	21.9	21.5
		FTI M1 [%] lo95	20.8	19.7	15.6	19.1	18.1	15.7	18.1	17.7	19.7	20.0
		FTI M2 [%] up95	17.5	16.6	16.3	12.0	13.6	11.9	13.2	12.3	14.2	14.7
		FTI M2 [%] lo95	15.7	10.3	6.8	10.5	11.4	8.7	11.9	10.4	12.5	13.2
		FTI M3 [%] up95	4.1	5.6	5.7	3.7	3.6	3.0	5.8	4.8	1.0	0.7
		FTI M3 [%] lo95	1.8	2.2	1.0	3.0	2.4	2.3	5.0	4.1	0.2	0.3
		FTI M6 [%] up95	17.6	21.3	23.5	17.3	12.0	12.8	12.5	13.5	15.3	16.5
		FTI M6 [%] lo95	13.3	16.8	7.2	14.5	10.0	10.2	11.0	11.3	9.5	12.6

FTI M7 [%]	up95	31.6	35.3	34.6	31.2	27.5	30.3	29.4	31.3	36.3	35.0
	lo95	30.3	29.3	20.8	29.1	26.2	28.3	27.0	29.2	33.9	33.2
FTI M8 [%]	up95	7.3	6.3	20.0	14.3	20.7	22.5	13.4	13.1	14.9	12.5
	lo95	4.0	2.8	5.6	12.3	18.2	16.2	10.3	10.7	9.7	9.4
FTI M9 [%]	up95	7.3	5.9	18.0	4.9	7.7	9.2	8.0	8.8	6.8	5.9
	lo95	4.5	2.8	0.0	3.1	6.2	6.7	6.5	7.6	4.0	4.5

Figure captions

Figure 1: Marker setup of SKIN and PIN markers involving the foot and tibia.

Figure 2: Division of the plantar pressure signal into 9 subareas as described in [16].
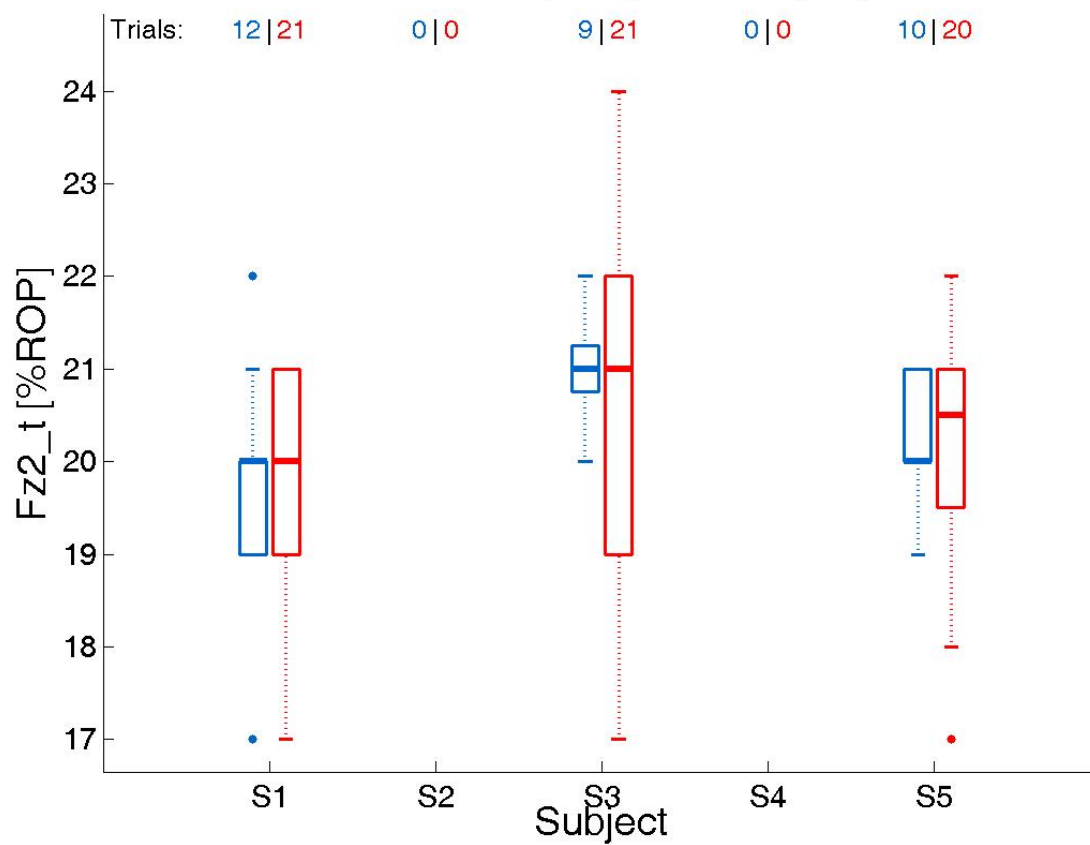
Figure 3: Boxplots of GRF variables. Small numbers above boxplots indicate the number of available trials per subject and condition.
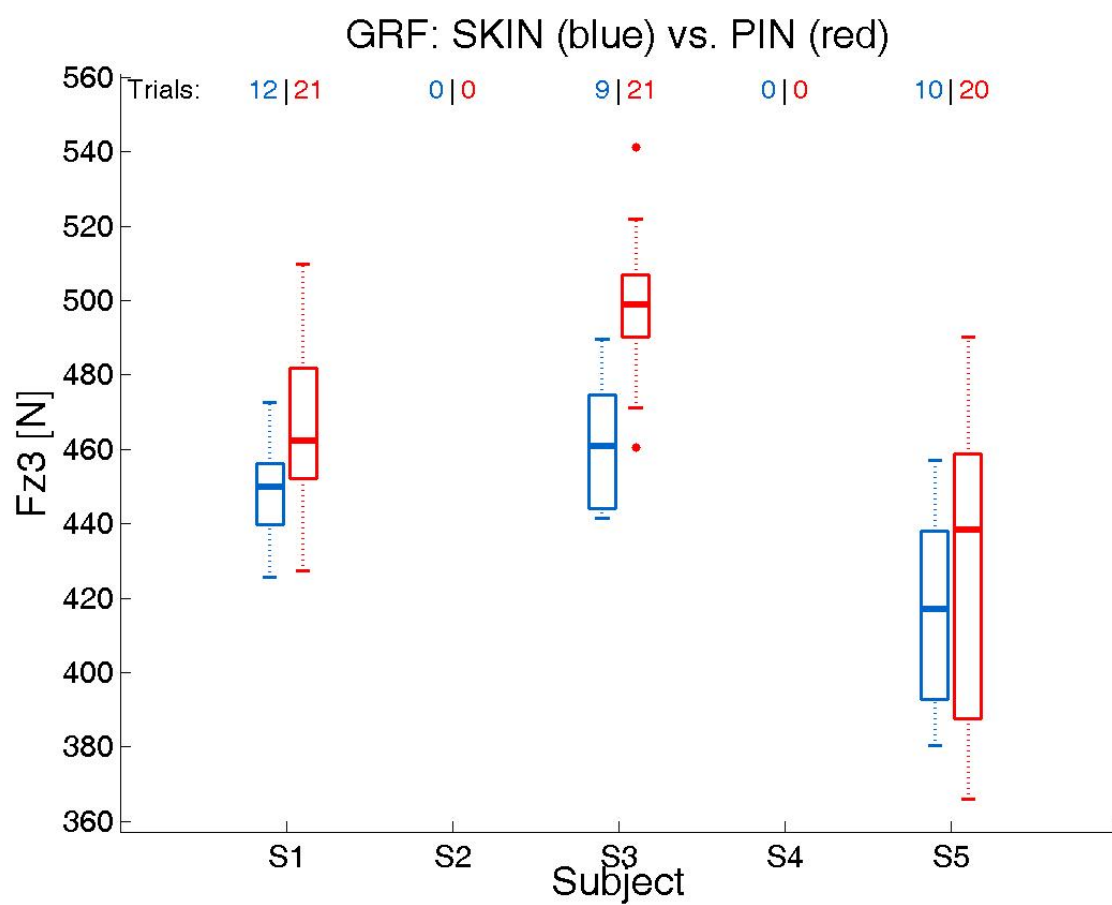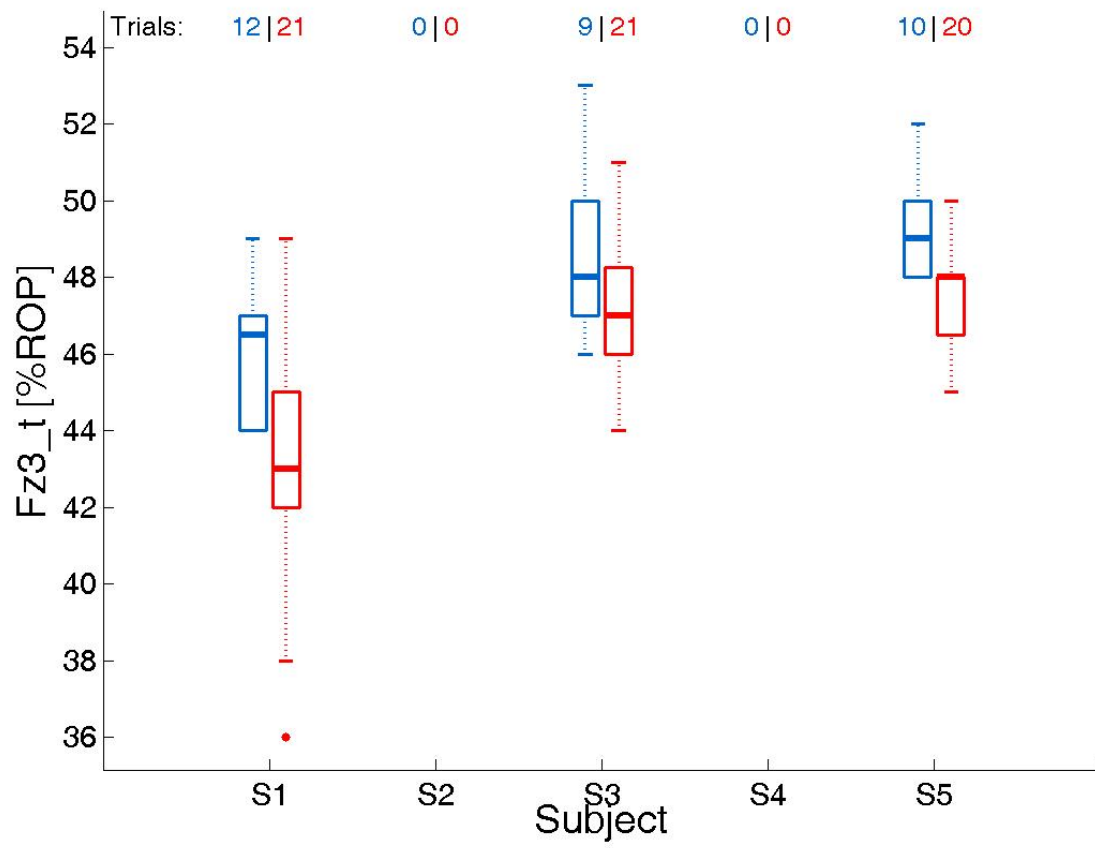
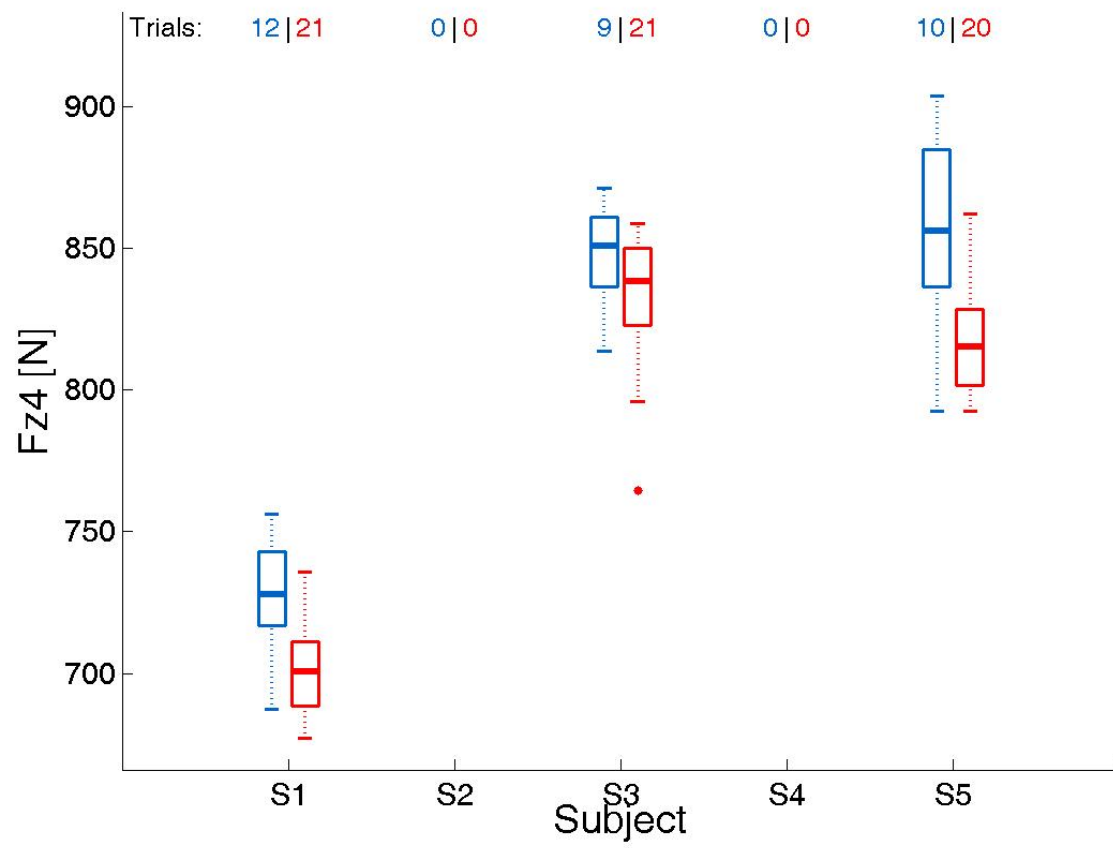GRF: SKIN (blue) vs. PIN (red)

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