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Simultaneous Bilateral Hip Joint Imaging at 7 Tesla using Fast Transmit B_1 Shimming Methods and Multichannel Transmission – A Feasibility Study

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Abstract

The objective of this study was to demonstrate the feasibility of simultaneous bilateral hip imaging at 7 Tesla. Hip joint MRI becomes clinically critical since recent advances have made hip arthroscopy an efficacious approach to treat a variety of early hip diseases. The success of these treatments requires a reliable and accurate diagnosis of intra-articular abnormalities at an early stage. Articular cartilage assessment is especially important to guide surgical decisions but is difficult to achieve with current MR methods. Because of gains in tissue contrast and spatial resolution reported at ultra high magnetic fields, there are strong expectations that imaging the hip joint at 7 Tesla will improve diagnostic accuracy. Furthermore, there is growing evidence that the majority of these hip abnormalities occur bilaterally, emphasizing the need for bilateral imaging.

However, obtaining high quality images in the human torso, in particular of both hips simultaneously, must overcome a major challenge arising from the damped traveling wave behavior of radio-frequency waves at 7 Tesla that leads to severe inhomogeneities in transmit B_1 (B_1^+) phase and magnitude, typically resulting in areas of low signal and low contrast, and consequently impairing use for clinical applications. In order to overcome this problem, a 16 channel stripline transceiver RF coil was used, together with a B_1 shimming algorithm aiming at maximizing B_1^+ in six regions of interest over the hips which were identified on axial scout images. Our successful results demonstrate that this approach effectively reduces inhomogeneities observed before B_1 shimming and provides high joint tissue contrast in both hips, while reducing the required RF power. Critical to this success was a fast, small flip angle B_1^+ calibration scan that permitted the computation of subject specific B_1 shimming solutions, a necessary step to account for large spatial variations in B_1^+ phase observed in different subjects.

INTRODUCTION

Assessing the hip joint with MR imaging becomes critically important since recent improvements in technique and instrumentation have made hip arthroscopy an efficacious way to treat a variety of early intraarticular hip problems (1). The enhanced awareness of diseases, such as femoroacetabular impingement (FAI) has led to its earlier recognition in adolescents and young adults with pre-arthritic and early arthritic disease (2). Accurate diagnosis not only of labral pathology but also of chondral lesions is important, because

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articular cartilage involvement has the most direct impact on surgical outcome (3,4). Such diagnosis however requires both high resolution and high contrast-to-noise ratio imaging (5,6). Furthermore, since there is growing evidence that the majority of these hip abnormalities occur bilaterally (7,8), often both hips should be imaged. Magnetic Resonance arthrography of a single hip using intraarticular gadolinium contrast agent at 1.5 or 3 Tesla (9–12) represents an improvement over conventional magnetic resonance imaging; however, it still has limitations when compared with direct observation, which are most pronounced when assessing articular cartilage (4).

Imaging at the ultrahigh field of 7 Tesla provides gains in signal-to-noise ratio (SNR) (13), greater parallel imaging performance (14), higher resolution, and potentially stronger MR contrast between joint tissues (e.g. (15–17)). However, a critical challenge to imaging the hips at 7 Tesla is the severe B_1 field inhomogeneities encountered as the hydrogen Larmor frequency reaches 300 MHz. At this high frequency, the radiofrequency (RF) wavelength in most tissues is about 13 cm, comparable or smaller than most human body part dimensions. As demonstrated in our early work, the resulting ‘damped traveling wave’ behavior (18) leads to large and spatially rapid phase variations (19). This phenomenon accounts for improved parallel imaging performance at ultrahigh fields (14), but leads to significant impediments on the RF transmit side (19) due to destructive interferences yielding highly inhomogeneous transmit B_1 (B_1^+) fields, especially in the human torso.

Here, we demonstrate bilateral human hip imaging at 7 Tesla. The approach utilizes multichannel transmit technology with a 16 channel independent RF transmitter coupled to a 16 channel transmit and receive (transceive) array coil. Such B_1 management techniques require calibration of the B_1^+ field of each individual transmit channel, potentially adding significant scan time to the study, which is undesirable in clinical practice. However, we also demonstrate the successful implementation in the hips at 7 Tesla of a fast, low SAR, B_1^+ calibration procedure. The results demonstrate that this multichannel transmit capability together with the fast B_1 shimming techniques efficiently addresses B_1 defects over both hips, providing satisfactory image quality and contrast homogeneity for bilateral hip joint imaging at 7 Tesla, while reducing specific absorption rate (SAR) levels.

MATERIALS AND METHODS

Eight healthy volunteers (four female and four male, ranging in age from 20 to 54 years, in weight from 56 to 79 kg and in height from 163 to 190 cm), who signed an informed consent form approved by our institutional review board, were scanned on a whole body 7 Tesla system (Magnex, Oxford, UK; Siemens, Erlangen, Germany). Two 8-elements Transceiver strip-line arrays (20,21) were positioned anterior and posterior to the pelvis (Fig. 1), with each coil element (15.3 cm $\times$ 1.27 cm) used for both RF transmission and reception. The 16 channels were powered with 16 $\times$ 1 kW CPC RF Amplifiers (CPC, Brentwood, NY), and each RF input phase was independently adjusted with a remotely controlled multi channel CPC RF phase/gain unit (CPC, Brentwood, NY), as previously detailed in (21). In addition to the standard MR console system monitoring the forward and reflected RF power on one channel, a similar 16-channel RF power monitoring system was developed in-house, continuously measuring the RF power delivered on each RF channel, while running averages were constantly updated over two time constants (10 s and 300 s). Whenever the measured average power exceeds predefined limits on any single RF channel, all RF amplifiers are instantly disabled. More details on the RF monitoring setup can be found in (21).

B_1 calibration

Complex $B_{1,k}^+$ maps of the 16 transmit (Tx) channels k [$k = 1, 2, \dots, 16$] were estimated using a recently introduced fast, low SAR multichannel method (22). The corresponding B_1 calibration data consisted of a series of 16 small flip angle gradient echo (GRE) images obtained while pulsing RF on one transmit channel k [$k = 1, 2, \dots, 16$] at a time. All data were sampled with the 16 receive channels. This interleaved multi-slice acquisition (total duration = 104 s) was obtained in three axial slices crossing both femoral heads, positioned on standard coronal scout images (see Fig. 2A). Measuring absolute flip angle values in-vivo in the femoral heads at 7T proved most of the time unstable and unreliable even when trying different B_1 mapping sequences. However, we inspected Turbo Spin Echo Proton Density weighted (PD-TSE) images acquired with different voltage and consistently found that image signal and contrast always kept increasing when applying a higher reference voltage (data not shown), up to the highest used voltage (90% of maximum allowed RF power) without reaching the maximum intensity that would occur if the flip angle used for excitation and refocusing pulses was actually close to 90 and 180 degree, respectively. Thus, given the inherently high dependence of T_2 -w image quality on flip angle, and knowing that no over-flipping was occurring, RF power was empirically calibrated in order to reach (but never exceed) 90% of the maximum allowed RF power during TSE sequences (RF monitoring will be described later).

B_1 shimming

Based on the B_1 calibration data, a transmit B_1 Multi Channel Interference Index¹ $B_1\text{MCII}$ was computed for each pixel, as described in (19,23), in order to estimate the ratio between the magnitude of the sum (MOS) of the 16 complex vectors $B_{1,k}^+$ (with $k = [1, 2, \dots, 16]$) and the sum of the magnitude (SOM) of the same vectors: $B_1\text{MCII} = \text{MOS} \div \text{SOM}$, with $\text{MOS} = |\sum_k (B_{1,k}^+)|$ and $\text{SOM} = \sum_k (|B_{1,k}^+|)$ (full details of this computation can be found in (19,23)). By definition, $B_1\text{MCII}$ ranges between 0 (fully destructive interferences) and 1 (fully constructive interferences), the latter case occurring when, and only when, the phase $\arg(B_{1,k}^+)$ of all channels k is identical within a considered pixel. At the beginning of each imaging session, the starting set of transmit phases, at the RF coil input, consisted of a phase increment of $360^\circ \div 16 = 22.5^\circ$ for every adjacent channel, following their geometrical order. To calculate new sets of transmit input phases for B_1 shimming, six disk-shaped ROIs were manually drawn (one ROI for each hip on each of the three slices) encompassing, and extending ~1 cm beyond the whole femoral head sections (see Fig. 2A). It was found that when the center area of these ROIs only included inner bone tissue (not joint), thus with very low GRE image SNR, better results were obtained by excluding the center part of these ROIs. Using the Matlab optimization toolbox (Matlab, Mathworks, Natick, MA), B_1 shimming solutions were obtained with an algorithm aiming at maximizing the value of $B_1\text{MCII}$ averaged over the ROIs, with a constraint on an upper limit for the ratio ($\max(|B_1^+|) - \min(|B_1^+|) \div \text{mean}(|B_1^+|)$) calculated over the same ROIs, where $B_1^+ = \sum_k B_{1,k}^+$. The 6 ROIs were treated as a single target in all computations. For technical reasons, only the relative phases, but not the relative magnitudes, of the 16 transmit channels were altered to obtain B_1 shimming solutions. The pixel-wise distribution of $B_1\text{MCII}$ through the ROIs was calculated before and after B_1 shimming. Small flip angle GRE images were obtained transmitting through all channels, before and after B_1 shimming, in the three axial slices used for B_1 calibration, to verify consistency between the resulting image and the corresponding image pattern predicted by the B_1 shimming algorithm. Note that, although

¹In order to avoid confusion, 'B1+ Multi Channel Interference Index' ($B_1\text{MCII}$) was used in the present manuscript, rather than 'Fraction of Available B1' as in (23), because 'Available B1' is often understood as describing the efficiency of the entire RF transmit chain (transmit B1 magnitude obtained in the sample per square root of watt delivered at the RF amplifier output), which was not investigated here.

using relative amplitude B_1 modulation between channels (additional to phase B_1 modulation) would increase the degrees of freedom for B_1 shimming, only B_1 phase modulation algorithms with constant B_1 amplitude were available throughout the current study in the home written interface allowing for fast MR data based B_1 shimming. In the current implementation, using an external computer for all Matlab computations, the time to complete the whole B_1 shimming procedure was about 12 minutes including: slice localization on scout images (~1 min 30s), test GRE image (~10 s), fast B1 mapping acquisition (1 min 44s), raw data transfer from console to external PC (~1 min), data formatting and B1 maps calculation (~3 min), drawing ROIs (~2 min), B1 shimming calculation (~20 s), transferring and applying B_1^+ phase solution to RF amplifiers (~20 s) and test image for B_1 shimming verification (~10 s).

SAR computation

In order to estimate maximum local SAR values, electromagnetic (EM) simulations were performed at 300 MHz using a finite difference time domain solver (XFDTD v6.5, Remcom, Pittsburg, PA), reproducing the design of our 16 channel RF stripline array loaded with the whole model of an adult human body ('Duke' model) with tissue properties (conductivity, permittivity and density) being defined for each $5 \times 5 \times 5 \text{ mm}^3$ voxel. After drawing ROI's on three axial slices through the femoral heads, the same algorithm utilized in-vivo was used to calculate a B_1 shimming solution with the simulated complex B_1^+ field maps.

The complex electric field maps were used to generate 10 g averaged SAR maps under two conditions: first, applying a uniform incremental phase distribution through all channels (22.5 degrees per channel), second, applying the ROI based B1 shimming solution. These SAR maps were normalized to an input RF power of 2.5 Watt for each of the 16 channels, corresponding to the maximum limit set for each RF channel on our real-time RF monitoring system. Thus, the maximum value of the SAR maps, reported in W/kg, is an estimation of the maximum local SAR that would occur in-vivo if continuously pulsing at the maximum RF power allowed on the system. For partial body SAR, in order to keep a conservative estimation, we assumed that the total input RF power ($16 \times 2.5 \text{ W} = 40 \text{ W}$) was converted into heat in the body, and we assumed that this heat was only distributed within the volume of tissues closest to the coil which, in simulations, receives 95% of the total SAR in a worse case scenario simulation (only constructive interferences between electric fields).

Hip Joint Imaging

In order to visually evaluate the impact of B1 shimming on bilateral hip joint imaging, multi-slice coronal oblique Turbo Spin Echo Proton Density weighted (PD-TSE) images were obtained at a medium spatial resolution ($0.78 \times 0.78 \times 4 \text{ mm}^3$, TE = 12 ms, TR = 3000ms, 2 averages, 9 slices with 1 mm gap, matrix 444×640 , 25% oversampling, turbofactor = 8), *before* and *after* B_1 shimming using same RF input power, with the product TSE sequence of our console that included one 3-lobe filtered sinc 90 degree excitation pulse, one 3-lobe filtered sinc 180 degree refocusing pulse, and multiple variable flip angle 3-lobe filtered sinc pulses. In some cases, additional fat suppressed (gaussian pulse) PD-TSE images were collected after B_1 shimming at a higher spatial resolution ($0.5 \times 0.5 \times 1.5 \text{ mm}^3$ before interpolation) in the coronal oblique plane with no slice gap (TE = 14 ms, TR = 3000 ms, 2 averages, 7 slices, matrix 704×1024 , 25% oversampling, turbofactor = 8).

RESULTS

B_1 shimming

The B_1 shimming procedure and related findings are summarized in Fig. 2, with the top row (Fig. 2A) showing the three axial slices and the six ROIs used for B_1 calibration and

shimming. Fig. 2B illustrates, for the center axial slice, the estimated $|B_{1,k}^+|$ maps for the 16 transmit channels as well as their relative phase maps (relative to channel #1). The very low B_1 MCII values measured *before* B_1 shimming through the ROIs, shown in Fig. 2C, indicate severe destructive B_1^+ interferences occurring in the hip joints. Not surprisingly, the lowest B_1 MCII values (white arrows) often correspond to spots of low signal intensity (dark spots) in GRE images *before* B_1 shimming (Fig. 2D). Large gains in constructive B_1^+ interferences (i.e. larger B_1 MCII values) in the ROIs used to compute the new RF phases can readily be seen in Fig. 2E after B_1 shimming, while Fig. 2F displays the corresponding disappearance of dark spots in GRE images (white arrows). In Fig. 2G, histograms of the pixel-wise distribution of B_1 MCII through the ROIs *before* and *after* B_1 shimming are shown. The mean value [$\pm$ std] of B_1 MCII increased by ~ 3.25 fold, from 0.20 [± 0.097] to 0.65 [± 0.131], while B_1 MCII inhomogeneity (std/mean) was reduced from 0.48 to 0.20. The sets of relative B_1 phase changes for the 16 channels, after B_1 shimming, are summarized in Table 1 for the eight subjects.

SAR computation

The maximum local 10 g averaged SAR value, for an input power at 100% of our maximum limit, was 4.10 W/kg with uniform incremental phases (CP-like mode) and of 9.70 W/kg after B_1 shimming, providing a reasonable margin of safety under the IEC guidelines (International Standard IEC 60601-2-33 2010), where local 10 g averaged SAR limits are defined as 10 W/kg in mode 1 and as 20 W/kg in mode 2. Note that our estimation can be considered as conservative as we did not subtract the reflective power from the monitored forward RF power and we ignored additional losses occurring in cables and connectors in the RF coil. For partial body SAR, the mass of tissue covered by the coil was estimated at about 21 kg (or 27% of the total body mass in the model), which, for an input of ($16 \times 2.5 = 40$ W), corresponds to 1.9 W/kg. This value is substantially smaller than the IEC limits for partial body SAR which, for 27% of body coverage by the coil, are of 7.85 W/kg in mode 1 and 8.38 W/kg in mode 2.

Impact on hip imaging

Fig. 3 demonstrates typical improvements obtained in PD-TSE image quality in the hip joints after B_1 shimming, while using the same RF power. As expected, the impact is even more dramatic than in GRE images because of the higher intrinsic sensitivity of Spin echo images to variations in B_1^+ magnitude. Areas virtually devoid of signal in the right hip *before* B_1 shimming became clearly visible and signal homogeneity improved through both hips. In all subjects, noticeable improvement in signal intensity and homogeneity was achieved in both hips using similar B_1 shimming procedures. In Fig. 4, non-enhanced tissue contrast in the hip at 7 Tesla can be better appreciated at a higher spatial resolution ($0.5 \times 0.5 \times 1.5$ mm³) after B_1 shimming. Depiction of important joint structures, including the articular cartilage and the labrum is feasible. An excellent contrast is observed in these 7 Tesla PD-TSE images between the joint fluid and the remainder of the tissues.

DISCUSSION

These initial results demonstrate that both hip joints can successfully be imaged simultaneously at 7 Tesla using multi channel transmit methodology. Using a rapid B_1^+ calibration was critical since phase and magnitude distributions of $B_{1,k}^+$ significantly vary between subjects, making it necessary to calibrate $B_{1,k}^+$ in 16 channels for each volunteer (23). Targeting two distant groups of ROIs to calculate B_1 shim solutions was also essential to successfully image both hips in a single acquisition. Besides the improvements in contrast and signal homogeneity, a fundamental advantage of increasing constructive interferences in transmit B_1 (i.e. achieving larger B_1 MCII values) is to lower the required level of RF power,

thereby potentially reducing SAR for a given excitation flip angle, as was demonstrated in the prostate at 7 Tesla in our previous work (23). For the particular data set shown in Figs. 2 and 3, with ~3.25 fold gain in B_1 MCII, ~10 times more RF power would have been required to obtain *without* B_1 shimming equivalent contrast to what was achieved *with* B_1 shimming. B_1 shimming solutions, however, also modify the electric fields resulting in a non linear dependence between RF power and SAR distribution. Furthermore, local SAR, rather than global SAR, typically represents the limiting safety factor at 7 Tesla. With regards to the SAR levels simulated for the current study, our results indicate that the B_1 shimming solution generates about 2.37 times higher maximum local SAR (and about 1.6 times higher global SAR) than the uniform incremental RF phases for a fixed level of input RF power. However, in order to achieve a given excitation flip angle, and despite higher SAR value per input watt, the B_1 shimming solution actually generate only ~23% of maximum local SAR (and ~16% of global SAR) when compared to the incremental RF phase setting, a direct benefit of the large increase in B_1 MCII. It should be noted that the B_1 shimming procedure selectively targeted the hip joints, explicitly ignoring the rest of the tissue, resulting in the formation of some dark areas outside the hip joints *after* B_1 shimming. By contrast, B_1 shimming solutions aiming at ‘flattening’ $|B_1^+|$ over an entire axial slice through the pelvis repeatedly yielded very low B_1 MCII values, due to the nature of B_1^+ distortions at 7 Tesla (19), making such solutions impractical. The robustness of the fast B_1 calibration proved to be superior when using left-right axis for phase encoding in GRE images to avoid vascular ghosting artifacts across the hips. Although this orientation requires a longer acquisition, we anticipate that 1 min 44 s for a calibration scan is suitable for clinical MR studies at 7 Tesla, especially considering that a direct implementation of all B_1 shimming tools on the console should substantially shorten other components of the procedure (data transfer and computations). Similarly to what our group had previously reported with B_1 shimming in the prostate (23), significant variations in B_1 shimming phase solutions were observed between volunteers (see Table 1), emphasizing the importance of acquiring subject dependent multi channel B_1 calibration. Overall, we found that using 3 axial slices to calculate B_1 shimming solutions was an acceptable tradeoff, providing satisfactory image quality over the entire hip joints while keeping the B_1 shimming procedure sufficiently short, whereas attempts made when using only a single central slice to calculate B_1 shimming solutions resulted in suboptimal and less stable B_1 shimming solutions (data not shown).

The very high intrinsic contrast observed at 7 Tesla in PD-TSE images within the articular cartilage and between the acetabular and femoral head articular cartilage, labrum and joint fluid holds strong promises. The data shown in Fig. 4 were obtained with a fairly long acquisition time of 11 min. However, neither parallel imaging nor partial Fourier acceleration was employed; these preliminary data provide solid grounds for further studies aiming at optimizing bilateral hip imaging protocols at 7 Tesla, at high spatial resolution, in order to enable clinical investigation of hip abnormalities at ultra high fields. It should be noted that the readable improvements that we report in TSE images were obtained despite the fact that we were using suboptimal flip angles (see methods); this strongly suggests that additional improvements in image quality could be obtained by further addressing SAR limitations, using, for example, lower SAR RF pulse design.

The transmit B_1 shimming methods presented in this paper would also be applicable to bilateral hip imaging at 3T, where B_1 inhomogeneities also occur even though at a lower scale, in the context of a growing number of scanners equipped with multi transmit capability.

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ABBREVIATIONS

FAI	femoroacetabular impingement
GRE	gradient recalled echo
MOS	magnitude of sum
MR	magnetic resonance
MRI	magnetic resonance imaging
PD-TSE	proton density turbo spin echo
RF	radio frequency
ROI	region of interest
SAR	specific absorption rate
SNR	signal-to-noise ratio
SOM	sum of magnitude
Tx	transmit

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Figure 1. One of the two 8-channel elements used in both transmit and receive mode (transceive) is shown. The patient's hip joints are positioned between the two 8-channel arrays with the femoral head in the center of the coil along the zdirection.

Figure 2.

Summary of B_1 Shim procedure. **A:** *Left:* coronal scout image showing the positioning of three axial slices (yellow lines) through the hips for B_1 shimming. *Right:* ROIs drawn over the three axial slices enclosing both hip heads. Smaller inner disks were excluded because of low SNR in central bone tissues. **B:** Estimated $B_{1,k}^+$ maps for the 16 channels shown for one of the three axial slices. *Left:* $B_{1,k}^+$ magnitude (arbitrary units). *Right:* B_1^+ phase (degrees) relative to the first channel. **C:** B_1 MCII (ranging between 0 and 1) within the ROIs (shown in A) before B_1 shimming. **D:** Corresponding GRE axial images before B_1 shimming. Note the dark areas crossing the hip heads where B_1 MCII is very low (arrows). **E:** B_1 MCII after B_1 shimming. **F:** GRE axial images after B_1 shimming. Note the gains in both B_1 MCII and signal intensity (arrows). **G:** Pixel-wise histograms, within the ROIs, of B_1 MCII before (left) and after (right) B_1 shimming.

Figure 3.

Multi slice coronal fat suppressed low resolution PD-TSE images ($0.78 \times 0.78 \times 4 \text{ mm}^3$) of a 52-year old female before (*upper row*) and after (*lower row*) B_1 shim, obtained with the same RF power and shown with the same gray scale. Note the correction of multiple dark areas after B_1 shim and the much improved signal homogeneity. (Absolute quantification would require measuring $|B_1|^2$ in the large flip angle regime and correcting for receive B_1 which is not shown for these images.)

Figure 4.

Multi slice coronal fat suppressed PD-TSE images after B_1 shim. Six oblique coronal slices are shown, depicting the femoroacetabular joint (C = articular cartilage, L = labrum, F = joint fluid). Both hips were simultaneously covered; images were individually cropped for display purpose. All images, including the enlarged and labeled image on the right was acquired at $0.5 \times 0.5 \times 1.5 \text{ mm}^3$ resolution, Relevant acquisition parameters: TR = 3000 ms, TE = 34 ms, no inter-slice gap). Acquisition time was 11 min in order to not exceed SAR limits; however, neither parallel imaging nor partial Fourier acceleration was used.

Table 1

Relative RF phase settings in degrees (range: 0° – 359°) for 16 channels (1–16) obtained *after* B1 shimming for eight volunteers (A–H). These phases are relative to a uniform incremental phase distribution of 22.5° per channel utilized as a starting set of RF phases. (*Before* B1 shimming the relative RF phase of each of the 16 channels equals zero, with a corresponding absolute phase distribution of $[(0 + k)^{\circ}, (22.5 + k)^{\circ}, (45 + k)^{\circ}, \dots, (315 + k)^{\circ}, (337.5 + k)^{\circ}]$ where the scalar constant k represents an absolute RF phase offset, uniform through all channels, that cannot be measured). There is no constraint on the overall average phase value.

	A	B	C	D	E	F	G	H
1	330	327	9	349	235	265	114	320
2	241	353	41	260	356	80	148	96
3	192	289	101	240	42	336	38	33
4	156	163	245	171	17	153	68	260
5	357	10	21	9	184	318	296	301
6	274	306	237	278	93	33	195	204
7	12	323	66	44	201	70	26	42
8	295	301	37	296	354	170	119	115
9	143	60	310	184	2	40	309	175
10	328	2	272	348	110	225	294	85
11	347	28	269	10	156	268	91	89
12	323	31	256	351	129	132	10	26
13	234	282	151	298	160	102	326	3
14	25	339	117	80	212	115	271	338
15	107	92	246	106	272	251	6	76
16	25	42	312	84	230	336	215	30