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Evaluation of a New $^1\text{H}/^{31}\text{P}$ Dual-Tuned Birdcage Coil for ^{31}P Spectroscopy

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Abstract

We introduce a new dual-tuned Hydrogen/Phosphorus ($^1\text{H}/^{31}\text{P}$) birdcage coil, referred to as split birdcage coil, and evaluate its performance using both simulations and magnetic resonance (MR) experiments on a 3 T MR scanner. The proposed coil simplifies the practical matters of tuning and matching, which makes the coil easily reproducible. Simulations were run with the Finite Difference in Time Domain (FDTD) method to evaluate the sensitivity and homogeneity of the magnetic field generated by the proposed ^1H coils. Following simulations, MR experiments were conducted using both a phantom and human thigh to compare the proposed design with a currently available commercial dual-tuned flexible surface coil, referred to as flex surface coil, for signal to noise ratio (SNR) as well as homogeneity for the ^{31}P coil. At regions deep within the human thigh, the split birdcage coil was able to acquire spectroscopic signal with a higher average SNR than the flex surface coil. For all regions except those close to the flex surface coil, the split birdcage coil matched or exceeded the performance of the flex surface coil.

Keywords

Dual-Tuned Birdcage; Spectroscopy; Magnetic Resonance Imaging; Chemical Shift Imaging

Introduction

Dual-tuned (e.g. $^1\text{H}/^{31}\text{P}$) birdcage coils have become a well known and widely used tool for magnetic resonance spectroscopy (MRS) [1-15]. They provide a uniform magnetic field inside the coil and can be made to fit most anatomies [1, 16-23]. MRS can be used for the study of an assortment of different metabolites in the human body. Oxidative metabolism indices, such as kinetic changes in phosphocreatine (PCr), are important indicators of biochemical processes [24, 25]. The development of dual-tuned coils has been explored with various nuclei [8], at different field strengths [15], and by studying animals [11], the human brain [1, 3, 12, 13], and skeletal muscles and neuronal synaptic development [26-33]. A two-in-one structure of such a coil solves practical issues that arise from having to use two separate coils or having to reposition the subject in the scanner. When spectral signal is localized to a specific region of tissue, it becomes very important to acquire a suitable anatomical image along with a multinuclear spectrum from the exact same tissue.

The ^{31}P MRS signal sensitivity is usually low due to low concentrations of the nucleus, a relatively small gyromagnetic ratio compared to ^1H , and magnetic field inhomogeneity from

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currently available commercial surface coils. Birdcage coils, on the other hand, give homogeneous magnetic fields and high efficiency at low specific absorption rates (SARs) [16-23, 34]. This point has been demonstrated in the past by simulating B_1 magnetic field maps in order to analyze homogeneity [18, 19, 21, 23] along with SAR calculations with a simulated model of the human body loaded in the coil [17, 20, 22].

Nonetheless, the practicality of building a dual-tuned birdcage coil is not trivial and can be a complex endeavor to approach due to many factors. Existing examples of dual-tuned birdcage coils include the 4-ring birdcage [1, 13], concentric birdcages [2, 8], and the alternating rung birdcage [9, 15]. All of these structures either share endrings or share enclosed sampling volumes.

In our study of human thigh muscle metabolism, the 4-ring design was initially chosen for its large sampling volume. However, it was found that construction of this coil required an iterative process of tuning the inner and outer sections, until the desired resonant frequency for each was reached. This process makes the coil very hard to reproduce. To overcome these difficulties, we propose the structure of the split birdcage coil for its large sampling volume, its use of inductive coupling between the inner and outer coils, and its full use of all of the legs in the coil. The split design uses coupling through the inner coil, similar to the 4-ring coil, but it separates the coils' direct connection to allow for the individual tuning of each coil.

In addition to various birdcage coils, dual-tuned surface coils have also been widely used primarily for their ease of use and high sensitivity close to the coil. We intend to find out whether the new design of birdcage coil can compete with the currently available commercial surface coil

In this work, the split birdcage coil's sensitivity, or B_1 field homogeneity, and SNR is presented with FDTD simulation. A magnetic resonance (MR) experiment was conducted to evaluate and compare the split birdcage coil and a commercial flex surface coil for their ability to localize and characterize a single muscle in the human thigh. All anatomical and spectroscopic signals were acquired from a phantom and a human volunteer using a GE 3 T magnet.

Materials and Methods

Simulations of the Split Birdcage Coil

The objective of the simulation was to evaluate the new split birdcage coil's ability to produce a homogeneous B_1 magnetic field for ^1H in the center of the coil. Simulations were run with a finite difference in time domain program, xFDTD (Remcom, State College, PA), to map the magnetic field. The simulated coil has three separate coils in a line that form three individual sections, an inner section and two outer sections, as shown in Figure 1. The whole structure was designed to be a low-pass low-pass (LP-LP) coil, which refers to capacitors placed on the legs of the inner and outer coils, so that the excitation sources could be placed easily and reliably on the coil legs in the simulation as the sources were limited to x-, y-, and z-orientations.

The optimization of the physical parameters of a 4-ring dual-tuned coil has been explored in the past [1, 12, 13]. However, the coils for this experiment were intended for imaging of the adult human thigh. The diameter, length of the inner coil, length of the outer coil, and coil separation were set at 23.45 cm, 20.79 cm, 4.51 cm, and 0.4 cm as seen in Figure 1, with consideration of optimal dimensions found by Duan et al [1].

The outer coils were tuned for ^1H imaging (127.72 MHz at 3 T) and the inner coil was tuned for ^{31}P MRS (51.7 MHz at 3 T). The overall grid size for the simulations was set to be larger than the size of the coil, and an adaptive grid was used that had a cell (i.e. voxel used in the simulation) size of 2.58 mm in the region of the coil and 21.66 mm far from the coil. Ten sinusoidal cycles of excitation were run to make sure that a steady state had been reached and a slice across the x-y or y-z plane was chosen at the same time point from each of two simulations. Rather than adjusting simulated capacitors at each leg of the outer birdcage coils, the coils were excited with a quadrature excitation method that used a constant (1 V), z-axis directed sinusoidal voltage source placed at the midpoint of each leg of the ^1H coils. The sources were phase shifted to form a sinusoidal current distribution, i.e. 0, 45, 90, 135, 180, 225, 270, and 315 degrees for the 8-leg coil. Capacitances placed on the inner coil were experimentally found on the bench to be 23.0 pF for the 8-leg coil. The resulting magnetic field is denoted with the real-valued coefficients of the x- and y-components of the transverse field C_{1x} and C_{1y} respectively. Then all of the current sources were phase shifted by 90° to form a cosinusoidal current distribution for a second run of the simulation, resulting in a magnetic field denoted with x- and y-component coefficients C_{2x} and C_{2y} . The magnitude of the total circularly polarized magnetic field resulting from the quadrature

excitation is given by $\frac{1}{2} \sqrt{(C_{1x} + C_{2y})^2 + (C_{2x} - C_{1y})^2}$ [35, 36].

Because the analysis was performed with free space loading, coil losses and tissue loading were not included in the simulation. According to the principle of reciprocity [37, 38], a transmit coil's transverse B_1 field per unit current is proportional to the signal acquired from the same coil during reception. Therefore, transverse B_1 field plots of each coil normalized to the center of the coil will provide an indicator of the coil's sensitivity profiles [37, 39]. Instead of presenting an absolute B_1 magnetic field, we normalized the B_1 fields of each coil to their centers in order to reflect the relative B_1 fields of these coils. The normalized transverse magnetic field magnitude was plotted in both the xy-plane (transverse plane) and the yz-plane (coronal plane).

Construction of the Split Birdcage Coil

The split birdcage coil was constructed on a 23.45 cm (9.25 inch) acrylic tube. Heat resistant tape and copper tape were laid down in the pattern of three separated 8-leg birdcages on a plastic film and wrapped around the acrylic tube. In contrast to the simulations of a LP-LP coil, a low-pass high-pass (LP-HP) design was chosen because the value of the capacitors used in the HP outer coils was larger than the equivalent LP birdcage with the same dimensions. A LP-HP design refers to the capacitors for the inner coil being placed on the legs and the capacitors for the outer coils being placed on the endrings. Also, it was experimentally easier to obtain symmetry of capacitance around the HP coil, because the capacitance could be more easily distributed. The prototype was built as an adaptation of this coil with an initial space of 0.4 cm between the inner coil and the high-pass outer coils. Additional components to our design included two quadrature couplers, matching circuits and phase shifters, and a protective plastic shell. The matching and phase shifting network are presented in Figure 1.

The inner ^{31}P coil was tuned with a capacitance at each leg of 21.2 pF. The inner coil was excited with quadrature excitation. Two equal length coaxial cables were used to deliver power to the coil at two capacitors that were 90 degrees from each other, yielding an improvement in SNR of the square root of 2 over linear excitation. A quadrature coupler (30 MHz to 88 MHz, Innovative Power Products, Holbrook, NY, USA) was used for this purpose. A T-shaped phase shifter was used for the inner ^{31}P coil to correct the phase shift to exactly 90 degrees. The two outer ^1H coils were each tuned to 127.75 MHz. Finally, two matching circuits were used to match each coil to 50 Ohms. Excitation on the ^1H coils gave

unloaded and loaded Q-factors of 125 and 90 respectively, while excitation on the ^{31}P coil gave unloaded and loaded Q-factors of 89 and 25 respectively.

The coil tuning was conducted using a large bottle (18 cm in diameter and 25 cm in length) filled with a 100 mM concentration of KH_2PO_4 as well as the human leg as a loading to ensure a realistic situation from the bench to the scanner. When the two outer coils were connected simultaneously, we observed the co-rotating and counter-rotating modes for each peak, similar to those reported in the 4-ring coil [13]. We tuned the co-rotating mode to the desired frequency.

Acquisition of Data

For comparison purposes, a flex surface coil (Clinical MR Solutions LLC., Brookfield, WI, USA) was used along with the split birdcage coil for data acquisition. The dimensions of the flex surface coil were 13 cm by 13 cm for the ^{31}P coil, and 20.5 cm by 22.5 cm for the two ^1H coils located on either side with 3.25 cm of overlap for each as seen in Figure 1. The total length of the flex surface coil was 41 cm and it could be wrapped around the subject. All data were acquired by using a GE 3 T HDx magnet (GE medical systems, Milwaukee, WI).

The anatomical image was acquired using a gradient recalled echo (GRE) sequence, followed by a 7×7 free induction decay chemical shift imaging (FID-CSI) sequence. The FID-CSI data was acquired after both auto- and manual-shimming to maximize signal intensity and peak shape.

For phantom imaging, parameters for anatomical image acquisition for both the split birdcage coil and the flex surface coil consisted of a repetition time (TR) of 350 ms, an echo time (TE) of 10 ms, a flip angle (FA) of 90 degrees, a field of view (FOV) of 18 cm, a slice thickness (STH) of 20 mm, and frequency and phase encoding matrix of 256×256 . While imaging the human leg, the parameters were the same for the flex coil, but the parameters for the split birdcage coil were TR of 100 ms, TE of 7 ms, FA of 90 degrees, FOV of 18 cm, and STH of 20 mm. This was done to keep the same in-plane and out-of-plane resolution for a fair comparison. The parameters for acquisition of FID-CSI data for both the split birdcage coil and the flex surface coil were: a TR of 3000 ms, a FA of 90 degrees, a FOV of 18 cm, a STH of 20 mm, and displayed grid dimensions of 7×7 . Four excitations were acquired resulting in a scan duration of 10 minutes.

The protocol was approved by the institutional review board of the University of Georgia, and all human data were acquired with informed consent. The flex surface coil was positioned such that the ^{31}P coil lay at the midpoint of the thigh, offset from the top towards the vastus lateralis muscle. The ^{31}P coil of the split birdcage coil was centered on the midpoint of the thigh. The FID CSI spectral voxels were approximately isotropic cubes measuring $2.57\times 2.57\times 2$ cm.

Data processing

All data processing was done with MATLAB (The MathWorks, Natick, MA). From the simulated magnetic field map, a region of interest (ROI) was selected within the boundaries of the FOV. For the axial view and the coronal view, the ROI is centered within the coil and is approximately a $6.33\text{ cm} \times 6.33\text{ cm}$ square, about one third the diameter of the phantom used in the experiment. A measure of the B_1 field sensitivity was displayed in Figure 2 with the normalized field maps from the simulations. Field map values greater than 35% of the field magnitude at the center were truncated so that field patterns in the center and those approaching the legs could be seen.

The experimental raw FID signal was loaded into MATLAB and then reconstructed with a two dimensional fast Fourier transform of the entire FOV in combination with a one-dimensional processing for a line broadening of 3 Hz. The SNR was calculated voxel-wise for the ^{31}P signal FID-CSI grids. The dominant ^{31}P peak (PCr) from each spectrum was about 400 data points away from the noise vector. The maximum value from each spectrum was divided by the standard deviation of the spectrum's noise to obtain the SNR.

Results

Homogeneity of the ^1H coil

Figure 2 shows the field maps that are normalized with respect to the center of their ROI obtained from simulation for a central axial slice (A) and a central coronal slice (B) for the split birdcage coil. For both slices, a region of homogeneity that is within 5% of falloff from the center is outlined by a black border as seen in Figure 2A and 2B. Also in Figure 2, because of the symmetry in the axial slice (i.e., x-y plane), only the x- and z-directed B_1 magnetic field profiles across the ROI are shown for the simulations (C and D). The sensitivity can be seen from the rate at which the field falls off. The area of the 5% homogenous region is 54.48% of the ROI in the axial slice. For the coronal slice, the region of homogeneity is 59.40%, shown in Figure 2B, by area of the ROI. The plots of the B_1 magnetic field profiles along x- and z-direction show that the field rises and falls in a smooth curve passing through the center of the normalized field.

Homogeneity of ^{31}P Signal

To compare the homogeneity, the data are displayed as a superposition of the FID-CSI grid on top of the anatomical image. Figure 3 displays the experimental results acquired from the phantom, with its magnitude of the spectral signal for each voxel in the 7×7 grid overlapped on the anatomical image. In Figure 3, it was observed from the phantom scans that the depth to which the flex surface coil can acquire ^{31}P signal is less than half of the 18 cm diameter phantom. The split birdcage coil, however, obtained signal uniformly from every voxel across the phantom including the voxel in the very center of the grid. As a comparison, the SNR was calculated at each voxel for both coil datasets and displayed in Table 1, which is organized in a way that directly correlates the position of the FID-CSI voxel and the position of the calculated SNR. For the flex surface coil, the SNR at areas close to the coil was ~ 275 and dropped off quickly with the distance from the coil. For the split birdcage coil, there were many voxels within the phantom that gave a SNR of 100. The best SNR value for the split birdcage was 119.01 while the best SNR for the flex surface coil was 298.63.

The comparison between the split birdcage and the flex surface coil for the human thigh is shown in Figure 4. The flex surface coil, when placed around the thigh orients the ^{31}P coil on top of the leg. The PCr peaks are visible for the flex surface coil at grid positions close to the coil. For the split birdcage, a PCr peak appears at every box which includes muscle. Quantitatively, Table 2 displays the calculated SNR from each voxel in the 7×7 FID-CSI grids. The highest SNR from the split birdcage is 174.45 while that of the flex surface coil is 305.73. The standard deviation was also calculated from the dataset containing the SNRs for the split birdcage coil and the flex surface coil on both the human thigh and the phantom. The deviation for the split birdcage coil for both the human thigh and the phantom was smaller than that of the flex surface coil. The deviation for the split birdcage on the human thigh was 61.0 compared to the flex surface coil's 77.5 and on the phantom, the deviation for the split birdcage coil was 34.3 compared to the flex surface coil's 97.2.

Discussion

The goal of this work was to introduce and demonstrate whether a split birdcage coil could effectively acquire signal from both the inner ^1H and outer ^{31}P coils. Previously, it has been shown that for a typical two-ring birdcage, the region of good homogeneity extends almost to the coil legs in the axial slice and almost to the end-rings in the coronal slice [11]. It has been shown that the split birdcage produces an effective circularly polarized field in the center of the coil. It is noted that the split birdcage coil also utilized the coupling between outer coils in order to generate a field in the center of the coil like the past 4-ring birdcage coils. This was seen simply from the appearance of corotating and counterrotating modes while tuning the coil on the bench.

The flex surface coil is designed to surround the ROI with two ^1H surface coils parallel to each other making up for its inability to acquire good signal at a distance away from the coil. The magnetic fields produced by the ^{31}P coil and the ^1H coils are orthogonal to each other. If the two ^1H coils do not completely surround the tissue, a dark spot will be produced. The dark spot is visible in Figure 3 and 4 on the far edge of the sample from the ^{31}P coil. However, the reason that this disadvantage is most often overlooked is due to the fact that surface coils are very easy to reproduce reliably. However, in order to acquire data from another muscle or muscle group, the coil must be shifted and repositioned [29]. A volume coil, that is fairly easy to reproduce and effective at multinuclear spectroscopy, has the advantage over the flex surface coil counterpart for various applications except attempting to acquire signal from tissues right under the coil.

By visual comparison of the images acquired with both the flex surface coil and the split birdcage coil, it is apparent that each coil has some dark spots. The flex surface coil has a large dark spot because the material to be imaged is too big for the flexible coil to fit around. The split birdcage also has symmetric dark spots on the outer edges. This is probably due to several factors, including a very tight-fitting birdcage coil around the human thigh, and the presence of the other leg just outside of the coil. These may cause misalignments of the linear modes with their connection points on the coil, especially for ^1H at 3 Tesla. Misalignment of the modes likely leads to the variation in intensity, which was observed in Figures 3 and 4. Nonetheless, neither of these dark spots hinders the function of the coil. The SNR for the image is comparable and, in the human thigh scans, all of the anatomy can be seen. That is what is required in order to map a particular ^{31}P spectrum to some localized region of the tissue. As seen in Figure 3 and 4, the ^{31}P signal drops off rather quickly for the flex coil on the phantom and slightly less quickly for the human thigh scan. The split birdcage on the other hand maintains a uniform ^{31}P signal across the subject, phantom or human thigh. By comparing the highest SNR from a voxel for the phantom scans, as seen in Table 1, it appears that the SNR from the split birdcage is less than half that of the flex surface coil. However, it also appears that the high SNR value continues to the center of the phantom for the split birdcage coil while the high SNR for the flex surface coil is limited to the area close to the coil. If the desired sampling volume is close to the surface, the flex surface coil will be better, however, the addition of any subcutaneous fat, with a thickness of 1 to 2 cm, will likely even the odds between the two coils.

Quantitatively speaking, Table 1 and 2 convey the numerical results for ^{31}P SNR calculations and show that the birdcage in fact has a smaller variation in ^{31}P peak intensity and therefore a more homogeneous B_1 magnetic field in the axial slice. Because of its effectiveness in acquiring signal for imaging and MRS like previous 4-ring birdcage coils, while at the same time exhibiting ease of construction and tuning, the split birdcage coil is a better choice for spectroscopy experiments.

The future of this research will be to consider human subjects with large layers of subcutaneous fat [40] which acts as a buffer between the muscle and the coil. As shown in this study, imaging of leg muscles is a challenging situation due to the use of tight-fitting coils and the close proximity of the two legs. Coils of this size may also radiate strongly in a high magnetic field, e.g. 3 Tesla. To mitigate these problems, a shield could be used around the coil. The split birdcage coil will potentially be essential for acquiring localized ^{31}P signal from the skeletal muscle of patients that might have a large layer of subcutaneous fat surrounding the thigh muscles.

Conclusion

A new split birdcage coil is introduced and evaluated in this study. The preliminary simulations showed the B_1 field sensitivity of the ^1H coils, and it was determined that the split birdcage coil would in fact operate under the same principles as previously researched 4-ring coils. The split birdcage coil was tested against a commercially available flex surface coil experimentally by acquiring ^{31}P spectra from a large 18 cm diameter phantom and a normal, healthy human thigh. The results showed that the split birdcage coil gave a more uniform ^{31}P signal from both the phantom and the human subject making it a much better choice for acquiring and localizing signal from a particular muscle in the tissue. This is a useful result because of the ease with which the split birdcage coil can be constructed. Having the three coils spaced apart instead of directly connected cut back the iterative tuning process and allowed for individual matching for each section without interference to the other coils.

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Split Birdcage Coil

Flexible Surface Coil

127.72MHz and 51.7MHz Layout

Matching Circuit (MC)

Phase Compensation Circuit (PCC)

Figure 1.

Schematic showing, the split birdcage coil (Top-Left), the flex surface coil (Top-Right), and the network of matching, splitting, and phase shift compensation to the coil (Bottom). The split birdcage was modeled in xFDTD and used in the simulations for ^1H field sensitivity. The constructed coil has a low-pass (LP) inner coil and high-pass (HP) outer coils as seen in this figure. For the proposed split birdcage coil, the matching circuits used were two component circuits composed of a capacitor or an inductor. The phase shifting circuits were T-shape circuits composed of two capacitors and one inductor.

Figure 2.

Simulation results from the split birdcage coil are displayed for an axial slice (A) and a coronal slice (B), followed by normalized magnetic field profiles across the FOV for the x-, y-, and z-directions. A and B show the magnitude of the normalized field with respect to the center of the ROI/FOV. The x-axes are displayed in units of cells from simulation results (0.21 cm/cell). The split birdcage coil is homogeneous across the ROI, highlighted by the rectangular box, with an area within 5% deviation encompassing 54.48% of the ROI for the axial slice and 59.40% of the ROI for the coronal slice. The thick black line on graphs (A) and (B) outlines the border of the region of 5% deviation from the field at the center of the ROI. Also, a consistent trend can be seen in the gentle rise to the center of the magnetic field profile in the transverse plane and a smooth fall to the center along the z-axis directed profiles of the coronal slices (D).

Figure 3.

This compares side by side the overlapped FID-CSI grid on top of the anatomical image for a 100 mM KH_2PO_4 phantom for both the split coil (Left) and the flex surface coil (Right). Note that the flex surface coil is over the top right of this image. The spectral peaks were acquired with the same digital, analog, and transmit gains. The ^{31}P voxel with the highest intensity peak (shown in the highlighted box) was selected and expanded above.

Figure 4.

Displayed in this figure are the overlapped FID-CSI on top of an anatomical image of the human thigh for the split coil (Left) and the flex surface coil (Right). Since the only source for PCr is the muscle, the partial volume effect can be seen at the boundaries of muscle to fat, bone, and blood vessels. There is reduced signal intensity when only part of the voxel is filled with signal producing sample. The ^{31}P voxel with the highest intensity PCr peak (shown in the highlighted box) was selected and an expanded view is shown above.

Table 1

presents numeric values for the SNR from each block in the corresponding FID-CSI grid from Figure 3. SNR table for A) the split birdcage coil and B) the flex surface coil, physically located at the top right of the FOV of the phantom.

A)									
8.59	15.39	66.69	44.55	40.80	26.25	9.49			
13.20	96.52	100.62	94.62	63.95	65.38	64.34			
82.68	102.83	102.70	90.69	70.01	83.77	53.47			
72.19	106.10	99.89	102.80	112.40	119.01	69.23			
44.08	90.39	86.74	75.68	83.10	89.29	56.83			
44.14	76.64	97.34	111.48	112.39	83.92	6.48			
3.87	21.17	56.94	68.95	46.96	6.53	5.51			

B)									
27.51	14.43	163.88	285.66	271.23	241.60	46.61			
40.76	88.07	242.05	263.93	280.58	298.63	262.25			
46.75	72.96	70.98	175.22	218.11	279.21	254.76			
6.92	33.32	70.78	106.86	137.18	116.19	51.86			
6.07	19.64	29.89	37.67	67.96	63.92	47.42			
8.77	19.35	53.43	45.07	73.02	20.41	34.18			
7.67	28.41	46.51	48.89	94.92	32.13	22.73			

Table 2

presents numeric values for the SNR from each block in the corresponding FID-CSI grid from Figure 4. SNR table for A) the split birdcage coil and B) the flex surface coil, physically located on top of the thigh.

A)							
4.99	8.83	5.58	7.38	8.60	8.32	6.10	
6.89	82.34	153.19	169.40	114.39	15.12	5.70	
51.23	170.17	117.48	139.11	146.47	86.33	6.58	
77.28	152.55	71.41	66.09	138.29	159.28	53.94	
7.42	86.28	113.00	109.51	174.45	112.48	5.71	
11.64	12.74	94.41	106.14	137.96	13.15	6.39	
5.42	7.10	14.57	17.69	10.88	6.29	4.32	

B)							
5.13	10.90	5.29	5.56	4.69	7.70	4.91	
7.95	74.60	176.93	161.37	192.78	53.10	7.12	
34.58	239.57	205.40	121.45	305.73	124.68	23.36	
17.75	165.51	213.06	40.22	143.93	53.69	19.36	
13.39	54.92	120.61	68.34	91.37	37.29	5.99	
4.55	14.98	22.45	37.25	35.06	5.89	6.61	
7.92	8.76	8.70	7.46	5.76	5.47	7.34	